Université de Montréal

Effect of modeling methods on the body and head-neck-trunk moments of inertia calculations in individuals of different morphology

by

Mohsen Damavandi

Département de Kinésiologie

Thesis submitted to the Faculté des études supérieures in partial fulfillment of the requirements for the degree of Philosophiae Doctor (Ph.D.) en sciences de l'activité physique

February 2008

Université de Montréal

Faculté des études supérieures

This thesis entitled:

Effect of modeling methods on the body and head-neck-trunk moments of inertia calculations in individuals of different morphology

presented by

Mohsen Damavandi

Was evaluated by the following examining committee:

Committee President:	Yvan Girardin, Ph.D.
Thesis Supervisor:	Paul Allard, Ph.D., P.Eng.
Internal Examiner:	François Prince, Ph.D.
External Examiner:	David Pearsall, Ph.D.
Representative of the Dean:	Jacques Bélair, Ph.D.

RÉSUMÉ

Le moment d'inertie segmentaire (MOI) peut être estimé d'après des données obtenues de cadavres ou de sujets vivants en utilisant différentes méthodes. Les études sur les cadavres ne peuvent être appliquées sur des populations spéciales, tandis que les études sur des sujets vivants sont onéreuses, invasives, ou dépendent de suppositions limitatives. La présente étude a été menée pour développer deux nouvelles méthodes in-vivo personnalisées, basées sur la dynamique inverse et l'approche du moment angulaire, pour estimer les valeurs de MOI du corps moins les segments des pieds et du complexe tête-cou-tronc. La méthode du moment angulaire fut préférée à la dynamique inverse pour estimer les valeurs de MOI du complexe tête-cou-tronc à cause de sa plus faible variabilité. Pour calculer les valeurs de MOI du complexe tête-cou-tronc sa masse et la localisation du centre de masse (COM) furent requis. Une nouvelle technique de plate-forme a d'abord été développée pour éviter l'utilisation des tables anthropométriques. Pour identifier la précision des méthodes proposées, leurs valeurs de MOI furent comparées à celle de de Leva, Hanavan et Jensen, chez des sujets ayant des morphologies normales, minces et obèses, pour vérifier si les méthodes se comportent de façon similaire à la méthode de de Leva.

Pour les valeurs de MOI du corps moins les pieds, aucune différence significative ne fut observée ($p<0.05$) entre les méthodes dans le groupe normal. Les valeurs de MOI des approches de dynamique inverse et de moment angulaire étaient en moyenne 17.3% plus élevées pour les participants obèses, et 13.3% plus faibles pour les participants minces ($p<0.05$) que les valeurs de MOI de la méthode de de Leva en référence aux trois axes principaux. Des coefficients de corrélation de Pearson ont montré que toutes les méthodes se sont comportées de façon similaire à la méthode de de Leva pour estimer les valeurs de MOI du corps moins les pieds dans les trois groupes morphologiques, à l'exception de la méthode de Jensen. Les méthodes de dynamique inverse et de moment angulaire se sont avérées sensibles aux valeurs de MOI pour diverses populations, spécialement pour les populations mines et obèses.

Des variations considérables ont été notées entre les méthodes pour estimer la masse et la position du COM du complexe tête-cou-tronc. Bien que des différences significatives aient été notées ($p<0.05$), la nouvelle technique de plate-forme a procuré des valeurs à l'intérieur de l'étendue des autres méthodes. Cette technique possédait en moyenne une corrélation plus faible (0.57) avec la méthode de de Leva pour la position du COM en comparaison aux autres méthodes. Ceci pourrait représenter une plus grande sensibilité de la nouvelle technique de plate-forme de force pour calculer la position du COM segmentaire dans différents groupes morphologiques.

Pour les valeurs de MOI du complexe tête-cou-tronc du groupe normal, les résultats étaient comparables bien que des différences ($p<0.05$) furent notées entre les méthodes de Hanavan et de de Leva. Pour les sujets minces et obèses, la méthode du moment angulaire a donné des valeurs MOI dans l'étendue des autres méthodes pour les trois axes principaux et s'est montrée sensible aux valeurs MOI pour différents types de morphologies. Cette méthode implique des calculs directs des valeurs du MOI des segments tout en évitant les limitations des autres méthodes.

En général, cette étude souligne l'importance de méthodes in-vivo personnalisées pour estimer les valeurs de MOI du corps et de ses segments dans une population comprenant différentes morphologies. Il est attendu que ce travail peut procurer plus de précision sur les paramètres d'inertie segmentaire, spécialement dans des populations minces et obèses.

Mots clés: Moment d'inertie, modèle de pendule inversé, dynamique inverse, moment angulaire, technique de plate-forme de force, morphologie corporelle, corps entier moins les pieds, complexe tête-cou-tronc

ABSTRACT

Segment moment of inertia (MOI) can be estimated from data obtained from cadavers or living individuals using different methods. The cadaver studies cannot be applied for special populations while living subject studies are expensive, invasive, or rely on some limiting assumptions. The present study was conducted to develop two novel in-vivo personalized methods based on inverse dynamics and angular momentum approaches to estimate MOI values of the body less the feet and head-neck-trunk segments. The angular momentum method was preferred over the inverse dynamics to estimate the head-neck-trunk's MOI values because of its less variability. To calculate the head-neck-trunk MOI values its mass and center of mass (COM) location are required. A new force-plate technique was first developed to avoid the use of anthropometric tables. To identify the accuracy of the proposed methods, their MOI values were compared to those of de Leva, Hanavan, and Jensen in subjects having a normal, lean, and obese morphology, and verified if the methods behaved similarly to the de Leva method.

For the body less the feet MOI values, no significant difference ($p<0.05$) was observed between the methods in normal group. The MOI values of the inverse dynamics and angular momentum approaches were in average 17.3% higher for obese, and 13.3% lower for lean participants ($p<0.05$) than those of the de Leva method about the three principal axes. Pearson coefficients of correlation showed all the methods behaved similarly with de Leva method to estimate the MOI values of the body less the feet in the three morphological groups except for Jensen's method. The inverse dynamics and angular momentum methods appeared to be sensitive to estimate the MOI values in various populations especially lean and obese.

Considerable variations were noted between the methods to estimate the head-neck-trunk's mass and COM position. Though significant differences were noted ($p<0.05$), the new force-plate technique provided values within the range of the other methods. This technique had in average a lower correlation (0.57) with the de Leva method for COM position compared to the other methods. This might represent more sensitivity of the new force-plate technique to calculate the segment's COM position in different morphological groups.

For the head-neck-trunk's MOI values in normal group, the results were comparatively similar though differences ($p<0.05$) were noted between the Hanavan and de Leva methods. For the lean and obese subjects, the angular momentum method gave the MOI values at the range of the other methods for the three principal axes and was noted to be sensitive to estimate the MOI values in different body morphologies. This method involves direct calculations of the segment's MOI values while avoiding the limitations of the other methods.

In General, this study underlines the importance of in-vivo personalized methods to estimate the MOI values of the body and its segments in population with different morphology. It is anticipated that this work can provide greater accuracy of segment inertial parameters especially in lean and obese populations.

Key words: Moment of inertia, inverse pendulum model, inverse dynamics, angular momentum, force-plate technique, body morphology, whole body less the feet, head-neck-trunk

TABLE CONTENTES

RÉSUMÉ....................iii

ABSTRACT....................v

LIST OF TABLES....................xii

LIST OF FIGURES....................xviii

LIST OF APPENDICES....................xxi

LIST OF ABBREVIATIONS....................xxii

DEDICATION....................xxiii

ACKNOWLEDGEMENT....................xxiv

Chapter 1

1. INTRODUCTION....................1

 1.1 Relationship between MOI values and body morphology....................3

 1.2 Current methods to estimate body segment moments of inertia....................4

 1.3 General objectives of the study....................9

Chapter 2

2. REVIEW OF LITERATURE....................11

 2.1 Review of body and its segments' MOI estimation methods....................11

  2.1.1 Cadaver studies....................11

  2.1.2 Geometrical modeling....................15

  2.1.3 Imaging techniques....................18

  2.1.4 Scanning techniques....................23

2.1.5 Oscillation techniques..28

2.2 MOI values of the whole body less the feet and head-neck-trunk in individuals of different morphology 34

2.3 Methods to estimate MOI values of the whole body less the feet and head-neck-trunk 36

2.4 Specific objectives of this thesis...39

Chapter 3

3. METHODS..41

3.1 Subjects..41

3.2 Moment of inertia of the whole body less the feet.........................43

3.2.1 Anthropometric, geometric, and photogrammetric methods........**43**

3.2.2 Inverse dynamics and angular momentum methods.................**50**

3.2.2.1 Data collection to calculate MOI of the whole body less the feet by the inverse dynamics and angular momentum methods.....**54**

3.3 Moment of inertia of the head-neck-trunk segment.....................59

3.3.1 Anthropometric, geometric, and photogrammetric methods........**59**

3.3.2 Angular momentum method..**60**

3.3.2.1 Calculations to estimate the segments' mass and COM position using a new force-plate technique..................................**60**

3.3.2.2 Data collection for the new force-plate technique to calculate segments' mass and COM position.............................**64**

3.3.2.3 Data collection for the angular momentum method to calculate head-neck-trunk segment's MOI values..........................67

3.4 Data analysis..**71**

Chapter 4

4. RESULTS..73

4.1 Whole body less the feet moment of inertia results......................**73**

4.1.1 Whole body less the feet MOI of the normal BMI range...........**74**

4.1.2 Whole body less the feet MOI of the lean group....................**76**

4.1.3 Whole body less the feet MOI of the obese group..................**78**

4.1.4 Pearson coefficients of correlation of the whole body less the feet MOI..**80**

4.2 Head-neck-trunk segment's mass and COM position...................**81**

4.2.1 Head-neck-trunk segment mass..**82**

4.2.2 Head-neck-trunk segment COM position...........................**84**

4.2.3 Pearson coefficients of correlation of the head-neck-trunk mass and COM position..**86**

4.3 Head-neck-trunk moment of inertia.......................................**87**

4.3.1 Head-neck-trunk moment of inertia of the normal BMI range......**88**

4.3.2 Head-neck-trunk moment of inertia of the lean group..............**89**

4.3.3 Head-neck-trunk moment of inertia of the obese group............**91**

4.3.4 Pearson coefficients of correlation of the head-neck-trunk MOI values..**92**

Chapter 5

5. DISCUSSION..94

5.1 Effect of the modeling methods to estimate the MOI values of the whole body less the feet in individuals of different body morphology..95

5.2 Comparison of the modeling methods to estimate the MOI values of the whole body less the feet in individuals of different body morphology..102

5.3 Effect of the modeling methods to estimate the mass, COM, and MOI values of head-neck-trunk in individuals of different body morphology..104

5.3.1 Effect of methods to estimate head-neck-trunk mass and COM position in individuals of different body morphology..............104

5.3.2 Effect of the modeling methods on the MOI values of head-neck-trunk in individuals of different body morphology..................111

5.4 Similarity of the modeling methods to estimate the MOI values of head-neck-trunk in individuals of different body morphology.......118

5.5 Limitations of the study..119

5.6 Future studies..122

Chapter 6

6. CONCLUSION..125

7. REFERENCES..129

LIST OF TABLES

Table 3.1. Means and (standard deviations) of age, mass, height and body mass index (BMI) of normal, lean and obese morphological groups along with the number of subjects (n) in each morphological group....................................**42**

Table 4.1. Means, standard deviations, and ranges of the moment of inertia ($kg.m^2$) for the whole body less the feet calculated at the center of mass of subjects with a normal BMI along with the % difference between de Leva (1996) method and the other methods. Data are presented for antero-posterior (AP), medio-lateral (ML), and longitudinal (LG) axes..**75**

Table 4.2. Means, standard deviations, and ranges of the moment of inertia ($kg.m^2$) for the whole body less the feet calculated at the center of mass of subjects with a lean BMI along with the % difference between de Leva (1996) method and the other methods. Data are presented for antero-posterior (AP), medio-lateral (ML), and longitudinal (LG) axes..**77**

Table 4.3. Means, standard deviations, and ranges of the moment of inertia ($kg.m^2$) for the whole body less the feet calculated at the center of mass of subjects with an obese BMI along with the % difference between de Leva (1996) method and the other methods. Data are presented for antero-posterior (AP), medio-lateral (ML), and longitudinal (LG) axes..**79**

Table 4.4. Pearson coefficients of correlation (*p* values) of the moment of inertia estimations between de Leva method and the other approaches for the normal, lean, and obese morphological groups about the antero-posterior (AP), medio-lateral (ML), and longitudinal (LG) axes..**81**

Table 4.5. Means, standard deviations (S.D.), and the ratios (S.D.) of head-neck-trunk mass (kg) as a percent of whole body mass (WBM) along with the % difference between the de Leva (1996) method and the other method. Data are presented for lean (L), normal (N), and obese (O) morphological groups.........**83**

Table 4.6. Means and standard deviations (S.D.) of trunk-head-neck center of mass location to the hip joint center (cm), along with the ratios (S.D.) as a % of segment length (SL) and the % difference between the de Leva (1996) method and the other methods. Data are presented for lean (L), normal (N), and obese (O) morphological groups..**85**

Table 4.7. Pearson coefficients of correlation (*p* values) of the head-neck-trunk segment's mass and center of mass location between de Leva method and the other approaches. Data are presented for the lean (L), normal (N), and obese (O) morphological groups..**87**

Table 4.8. Means and standard deviations of head-neck-trunk moments of inertia (kg/m^2) of subjects with a normal BMI, along with radii of gyration (S.D.) as a %

of the segment length (SL) and the % difference between the de Leva (1996) method and the other methods. Data are presented for antero-posterior (AP), medio-lateral (ML), and longitudinal (LG) axes....................................**89**

Table 4.9. Means and standard deviations of head-neck-trunk moments of inertia (kg/m^2) of subjects with a lean BMI, along with radii of gyration (S.D.) as a % of the segment length (SL) and the % difference between the de Leva (1996) method and the other methods. Data are presented for antero-posterior (AP), medio-lateral (ML), and longitudinal (LG) axes...**90**

Table 4.10. Means and standard deviations of head-neck-trunk moments of inertia (kg/m^2) of subjects with an obese BMI, along with radii of gyration (S.D.) as a % of the segment length (SL) and the % difference between the de Leva (1996) method and the other methods. Data are presented for antero-posterior (AP), medio-lateral (ML), and longitudinal (LG) axes....................................**92**

Table 4.11. Pearson coefficients of correlation (*p* values) of the head-neck-trunk moment of inertia estimations between de Leva method and the other approaches for the normal, lean, and obese morphological groups about the antero-posterior (AP), medio-lateral (ML), and longitudinal (LG) axes.............................**93**

Table B.1. Means, standard deviations (S.D.), and the ratios (S.D.) of upper arm mass (kg) as a percent of whole body mass (WBM) along with the % difference between the de Leva (1996) method and the other method. Data are presented for lean (L), normal weight (N), and obese (O) subjects................................**148**

Table B.2. Means and standard deviations (S.D.) of upper arm center of mass location to the shoulder joint center (cm), along with the rations (S.D.) as a % of segment length (SL) and the % difference between the de Leva (1996) method and the other methods. Data are presented for lean (L), normal weight (N), and obese (O) subjects...**149**

Table B.3. Means, standard deviations (S.D.), and the ratios (S.D.) of forearm and hand mass (kg) as a percent of whole body mass (WBM) along with the % difference between the de Leva (1996) method and the other method. Data are presented for lean (L), normal weight (N), and obese (O) subjects................**150**

Table B.4. Means and standard deviations (S.D.) of forearm and hand center of mass location to the elbow joint center (cm), along with the rations (S.D.) as a % of segment length (SL) and the % difference between the de Leva (1996) method and the other methods. Data are presented for lean (L), normal weight (N), and obese (O) subjects..**151**

Table B.5. Means, standard deviations (S.D.), and the ratios (S.D.) of total arm mass (kg) as a percent of whole body mass (WBM) along with the % difference between the de Leva (1996) method and the other method. Data are presented for lean (L), normal weight (N), and obese (O) subjects................................152

Table B.6. Means and standard deviations (S.D.) of total arm center of mass location to the shoulder joint center (cm), along with the rations (S.D.) as a % of segment length (SL) and the % difference between the de Leva (1996) method and the other methods. Data are presented for lean (L), normal weight (N), and obese (O) subjects...153

Table B.7. Means, standard deviations (S.D.), and the ratios (S.D.) of thigh mass (kg) as a percent of whole body mass (WBM) along with the % difference between the de Leva (1996) method and the other method. Data are presented for lean (L), normal weight (N), and obese (O) subjects...154

Table B.8. Means and standard deviations (S.D.) of thigh center of mass location to the hip joint center (cm), along with the rations (S.D.) as a % of segment length (SL) and the % difference between the de Leva (1996) method and the other methods. Data are presented for lean (L), normal weight (N), and obese (O) subjects..155

Table B.9. Means, standard deviations (S.D.), and the ratios (S.D.) of leg and foot mass (kg) as a percent of whole body mass (WBM) along with the % difference between the de Leva (1996) method and the other method. Data are presented for lean (L), normal weight (N), and obese (O) subjects................................**156**

Table B.10. Means and standard deviations (S.D.) of leg and foot center of mass location to the knee joint center (cm), along with the rations (S.D.) as a % of segment length (SL) and the % difference between the de Leva (1996) method and the other methods. Data are presented for lean (L), normal weight (N), and obese (O) subjects..**157**

Table B.11. Means, standard deviations (S.D.), and the ratios (S.D.) of total leg mass (kg) as a percent of whole body mass (WBM) along with the % difference between the de Leva (1996) method and the other method. Data are presented for lean (L), normal weight (N), and obese (O) subjects................................**158**

Table B.12. Means and standard deviations (S.D.) of total leg center of mass location to the hip joint center (cm), along with the rations (S.D.) as a % of segment length (SL) and the % difference between the de Leva (1996) method and the other methods. Data are presented for lean (L), normal weight (N), and obese (O) subjects..**159**

LIST OF FIGURES

Figure 2.1. Lateral and frontal views of 17-segment of geometrical model (Adapted from Hatze, 1980). The shapes of the segments and local (segment fixed) coordinate systems are also shown..17

Figure 2.2. Lateral and frontal views of 16-segment photogrammetry model (Adapted from Jensen, 1978). X, Y, and Z indicate antero-posterior, medio-lateral, and longitudinal directions of the body..19

Figure 2.3. Recording positions, A) anterior view, B) lateral view (left side), C) coronal view. (Adapted from Baca, 1996)..22

Figure 2.4. Example of a whole body scan of human using a dual energy x-ray absorptiometer. Attenuation coefficients based on x-ray absorption values are recorded in elements of x-ray detector, resulting in samples of mass per whole body scan. (Adapted from Durkin et al. 2002)..26

Figure 2.5. Illustration of the measurement device to measure moments of inertia of the whole body (Adapted from Matsuo et al., 1995)................................30

Figure 2.6. The measurement postures of moment of inertia (Adapted from Matsuo et al., 1995)..30

Figure 2.7. Determination of the moment of inertia of the human body by the pendulum method (Adapted from Zatsiorsky, 2002)..................................**32**

Figure 3.1. Hanavan's (1964) geometrical model of the body. (Adapted from Robertson et al., 2004). The body segments are considered as geometric shapes and numbered as defined by Hanavan (1964)...**46**

Figure 3.2. A) Frontal, B) Lateral views of a subject for the photographs. The pictures were taken simultaneously by two digital cameras. The cameras-subject distance was 4.5 meters...**48**

Figure 3.3. Inverse pendulum model of the human body representing the COM oscillations about a point (*C*) lying midway between the ankles. Symbols are described in the text..**51**

Figure 3.4. Illustration of the three self-imposed oscillations of the whole body less the feet on: A) antero-posterior, B) medio-lateral, and C) longitudinal axes through a mid-point between the ankles...**57**

Figure 3.5. Illustration of the change in center of pressure as related to the change in segmental center of mass location during quite standing on the force-plate.....**62**

Figure 3.6. Illustration of the change in moments as related to the change in segmental center of mass location during lying on the reaction board..............**64**

Figure 3.7. Illustration of the markers put over the right upper and lower limbs calculated COM positions, malleoli, right humeral head, and greater trochanters to determine the horizontal positions of the upper and lower limbs.....................**68**

Figure 3.8. Illustration of the three self-imposed oscillations of the upper and lower extremities about: A and D) antero-posterior, B and E) medio-lateral, and C and F) longitudinal axes through the related proximal joint centers..................**70**

LIST OF APPENDICES

Appendix A: Research approval by Sainte-Justine Hospital Ethics Committee……
……………………………………………………………………**147**

Appendix B: Mass and COM positions of the upper and lower limbs and their sub-segments……………………………………………………………**148**

LIST OF ABREVIATIONS

2-D	Two-dimensional
3-D	Three-dimension
ANOVA	Analysis of variance
AP	Anterior-posterior axis
BMI	Body mass index
cm	Centimeter
COM	Center of mass
COP	Centers of pressure
CT	Computerized tomography
DEXA	Dual energy x-ray absorptiometry
Hz	Hertz
LG	Longitudinal axis
ML	Medio-lateral axis
mm	Millimeter
MOI	Moments of inertia
MRI	Magnetic resonance imaging
s	Second
SIP	Segment inertial parameters

I would like to dedicate this thesis, with love, to my wife and daughter.

ACKNOWLEDGEMENTS

This thesis would not have been possible without the support and generosity of some very special people who devoted their time and energy to provide me with their valuable contributions.

I wish to extend my sincere thanks to my supervisor, Dr. Paul Allard for giving me the opportunity to study in his laboratory. His belief in me always has been an inspiration, and his wisdom, patience and challenging supervision of this thesis, have been extremely indispensable. I am thankful to have had the chance to be his student; his encouragement has prepared me for my future endeavors.

I would like to acknowledge the Ministry of Science, Research and Technology of the Islamic Republic of Iran for financial support during my studies to do my degree in Canada.

Many thanks also to my co-workers and friends, especially Dr. Nader Farahpour, Dr. Franck Barbier, and Dr. Julien Leboucher for their technical assistance.

Finally, I would like to acknowledge my debt to my dear wife for her tolerance, take care of our daughter, Nahid, and her understanding during my studies.

Chapter 1

1. INTRODUCTION

Biomechanical analysis of the human body requires accurate segment inertial parameters that include mass, center of mass (COM) location, and moments of inertia (MOI). MOI values are necessary to calculate joint muscle moments during gait, sport activities, etc. Most often these values are estimated from anthropometric tables obtained from a limited number of elderly male cadavers (Dempster, 1955; Chandler et al., 1975). Though these values can be applied to most able-bodied adults, they do not represent accurately the adolescent population (Reid and Jensen, 1990) or individuals with distinct morphologies like children (Jensen, 1986), the obese (Pearsall et al., 1994), and scoliotic populations to name a few. In these special populations body segment proportions are different from those found in the anthropometric tables (Zatsiorsky, 2002). Improving the specificity of the MOI estimation techniques for individual body type, gender, and age groups could reduce errors in biomechanical calculations (Jensen, 1993).

The limitations in using cadaver-based methods to estimate MOI values led to alternative techniques where measures were taken from living subjects. These techniques include among others geometrical modeling (Hanavan, 1964), photogrammetry (Herron et al., 1974; Jensen, 1978; McConville et al., 1980), gamma mass scanning (Zatsiorsky and Seluyanov, 1983, 1985), computerized tomography (CT) imaging (Ackland et al., 1988), magnetic resonance imaging

(MRI) (Cheng et al., 2000), and dual energy x-ray absorptiometry (DEXA) (Durkin and Dowling, 2003). However, these methods have also some limitations. For instance, in the photogrammetric method a uniformed density of the mass distribution in each body segment is assumed. This assumption could lead to an overestimation of the mass and MOI values of some of the body segments such as the trunk (Jensen and Fletcher, 1994; Plagenhoef et al., 1983). Gamma mass scanning technique (Zatsiorsky and Seluyanov, 1983, 1985) has the ability to measure mass distribution within each body segment. However, Reid and Jensen (1990) reported that a wide range of differences in segment inertial parameters were noted between various populations. Though the radiation techniques provide accurate MOI measurements (Pearsall et al., 1996) these are expensive and rely on the use of well-trained operators. In addition, some radiation techniques like DEXA are able to provide MOI values only in the frontal plane (Durkin and Dowling, 2003). Personalized in-vivo methods where no assumption is made on segments' shape and density could be more appropriate to estimate the MOI values in a clinical or biomechanical laboratory setting.

In this chapter, the relationship between the MOI values and the body morphology will be addressed. This will be followed by a description of the current methods to estimate the MOI values for the whole body and particularly for the trunk in relation to body morphology. The need for new personalized in-vivo methods to estimate MOI values of the body which can be applied for any kind of population will be outlined as well. Finally, the general objectives of this thesis will be presented.

1.1 Relationship between MOI values and body morphology

The moment of inertia of a body segment depends on its mass and on the distribution of mass within the body with respect to the axis of rotation. The distribution of mass about the given axis is represented by the radius of gyration. Pearsall et al. (1996) and Durkin and Dowling (2003) reported that segments' mass expressed as a percentage of the total body mass, and radii of gyration of body segments are different in people of various body morphologies. With respect to the whole body, MOI values strongly depend on body size. Body mass and MOI values are approximately proportional to the subject's height to the third and fifth power, respectively (Zatsiorsky, 2002). Consequently small changes in body size could result in large changes in the MOI values. This can be appreciated in growing children and subjects of different morphologies.

MOI values vary within and between different populations, such as males and females, different races, and sports groups (Reid and Jensen, 1990). Considerable changes in the inertial parameters occur across the life span as individuals grow (Jensen, 1986), develop, and age (Jensen, 1994; Pavol et al., 2002). Current MOI methods have been developed mostly based on a specific sample (e.g., old male cadavers, young adults Caucasian). Therefore, these methods produce inaccuracies when applied to people with different morphologies than those of the original studies (Ganley and Powers, 2004). Body mass index

(BMI) is often used to classify individuals into lean, normal and obese types. To our knowledge, the effect of the MOI methods (e.g., de Leva, 1996; Hanavan, 1964) in individuals classified according to their BMI has not been investigated. These effects could provide insight into the sensitivity of the MOI models applied to population of different morphology.

1.2 Current methods to estimate body segment moments of inertia

The common methods to estimate the MOI values of the body and its segments and their effect on the accuracy of the results in population with different morphology will be briefly described to highlight their capabilities and limitations. These methods are well-known and have been applied frequently in biomechanical evaluations such as gait studies (Pearsall and Costigan, 1999; Rao et al., 2006).

Attempts to provide data on body segments' MOI values began with the onset of cadaver studies. These studies consisted of sectioning cadavers into segments and measuring the inertial parameters directly (Dempster, 1955, Clauser et al., 1969; Chandler et al., 1979). The moment of inertia of each body segment was calculated using a pendulum technique. Then, regression equations were developed to estimate MOI values of individual body segments. Since the radii of gyration are derived from a small sample of elderly (e.g., eight cadavers), male, and Caucasians, the findings should be restricted to a similar population. The assumption that inertial parameters derived from cadavers vary little from the living subjects has been questioned by Clarys et al. (1984), Martin et al. (1989), and Reid (1984). Achard et al. (2006) underlined that MOI estimates based on cadaver studies can be a source of error on kinetics analysis of human performance of people with distinct morphology particularly in the obese subjects.

The shortcomings in the MOI estimations using cadaver-based methods led to alternative techniques for measuring inertial properties more directly from living subjects. The use of living subjects offers the possibility of sampling populations more adequately by increasing the size of the sample. These methods were divided into geometric, photogrammetry, scanning and imaging techniques, and oscillation techniques.

The geometrical modeling technique is based on the representation of a segment or its components by standard geometric shapes of known density. It is assumed that each segment is a single homogeneous solid such as a right elliptical cylinder or a frustum of right circular cone (Whitsett, 1963; Hanavan, 1964). Dimensions for the body segment shapes are based on anthropometric measures taken on the subject while the segments' mass is estimated from regression equations based on cadaver studies (Barter, 1957). By taking additional anthropometric measures, such as mid-thigh circumference and knee diameter, Hanavan (1964) determined the principal moments of inertia. The accuracy of these methods has been questioned because of the simple geometric shapes and uniform mass distribution assumptions. The single homogeneous solid assumption fails to take into consideration the shape fluctuations throughout the length of each segment, a problem recognized by Hanavan in his report (1964, p. 39). Thus, this model can affect segment inertial parameters especially for segments with complex contours (Rao et al., 2006).

Given the inaccuracies associated with the identification of segments as simple geometric shapes (e.g., right elliptical cylinders), photogrammetric methods were developed. To individualize body segment moment of inertia values, Jensen (1978) developed a photogrammetric method in which segments are sectioned into elliptical discs of 20 mm width. This method takes into account the differences of body segments' volume and shape in individuals with different body morphology (e.g., lean, obese) but still makes the assumption that the segment densities are known. Furthermore, the photogrammetric method was found to overestimate body segments' volume (Kaleps et al., 1984) and the principal MOI values (Hatze, 1980). Consequently, the inertial parameters obtained from photogrammetric method should be applied with caution.

Another approach to determine the segments' MOI values involves scanning the living body with various radiation techniques. For instance, Zatsiorsky and Seluyanov (1983, 1985) presented data from an extensive study on the body segment parameters of both college-aged Caucasian males and females. They used gamma mass scanning to quantify the density of incremental slices of each segment. These results were then applied to compute segmental mass distribution. This method enabled estimations of the mass, COM, and principal MOI values in three-dimension (3-D) of the body segments. Since the inertial parameters were obtained from a young adult Caucasians population they could be misleading to individuals with different body morphology.

Other radiation techniques were developed to quantify segments' MOI values. These include CT imaging (Huang and Suarez, 1983; Reid, 1984), MRI (Mungiole and Martin, 1990; Cheng et al., 2000), and DEXA (Durkin et al., 2002). Though these latter approaches have the advantage of measuring the tissue distribution within in-vivo, they were not used widely owing to the health risks of radiation exposure and expensive instrumentation. Dual energy x-ray absorptiometry (Durkin and Dowling, 2003) was also limited to calculate the MOI values of the segments in frontal plane. Thus, they have limited application in routine clinical assessments of segment inertial parameters.

Personalized methods were mostly developed to measure MOI values of the extremities. These techniques involve an oscillation technique (Hatze, 1975) and quick-release method (Drillis et al., 1964). Both techniques require that the body part be set into oscillation while the muscles are relaxed so that they do not influence the damped oscillations and acceleration of the oscillated limb. In the oscillation technique a segment oscillates by means of an instrumented spring, while quick-release method assumes the acceleration of a rapidly accelerated segment is affected only by the segment's rotational inertia. Euler's equation of motion is then used to estimate the oscillated limb's moment of inertia. While these techniques can be applied to calculate the MOI values of any kind of population, their application is limited only to the upper and lower limbs. Therefore, these techniques cannot be used to calculate the MOI values of the trunk segment.

Personalized in-vivo methods where no assumption is made on segments' shape and mass density could provide greater accuracy to estimate the MOI values in subjects having different body morphologies. Two new methods based on 3-D inverse pendulum model are proposed in this thesis to estimate the whole body less the feet MOI values. The first method is based on Barbier et al. (2003) where the human body is modeled as a 3-D inverse pendulum representing a point mass oscillating about the ankles (Morasso et al., 1999; Brenière, 1996). In Barbier et al. (2003) the 3-D excursion of the COM is estimated from external forces, ankle muscle moments, and inertial properties. Conversely, if the COM angular accelerations are determined from videography, then the inertial properties of the oscillating whole body can be estimated for each individual. The second method applies the angular momentum equation. It consists of tracking the body during self-imposed oscillations about the ankles by means of a video-based system and force-plate. Then, the angular momentum of the body is calculated from video data and the integration of the moments obtained by the force-plate. Since the feet are fixed to the ground during the oscillations about the three principal axes of rotation, these two methods can be applied to estimate the whole body MOI values but the feet.

Since the moments of inertia of the individual body segments may do not represent the effect of the used methods on the estimated values, these new methods were first tested for the whole body less the feet. This could serve as a means of gaining insight into the sensitivity of the de Leva (1996), Hanavan (1964), and Jensen (1978) methods in subjects with different body morphologies.

Trunk represents a segment with the greatest variation of reported inertial parameters (Pearsall et al., 1994; Zatsiorsky, 2002). For instance, trunk mass/body mass ratio varies from 42.2% (Pearsall et al., 1994) to 52.4% (Chandler et al., 1975). Even when the sample is homogenous (e.g., adult men), the range can be as large as 12.2%, from 35.8% to 48.0% (Pearsall et al., 1994). This discrepancy is due to the use of various measurement techniques, subject samples (cadaver *versus* living), and segment definitions. Employing inertial parameters ratios derived from a sample population to estimate inertial properties of distinctly different populations would lead to substantial errors. In selecting a method to estimate segments' MOI values, the age, sex, and body morphology of the sample should be considered.

1.3 General objectives of the study

The general objective of this thesis was to test the ability of two new personalized in-vivo methods (inverse dynamics and angular momentum) to estimate MOI values compared to those obtained by de Leva (1996), Hanavan (1964), and Jensen (1978) methods. Furthermore, this thesis aimed to test the effect of these methods on population with different body morphology using BMI. A new technique was developed to estimate the MOI values of the head-neck-trunk segment and compared to the de Leva (1996), Hanavan (1964), and Jensen (1978) methods in subjects of different morphology.

Chapter 2

2. REVIEW OF LITERATURE

This chapter reviews the most commonly used methods for the estimation of the moments of inertia of body and its segments with the objective of presenting their advantages and disadvantages. Next, the need to estimate the MOI values of the whole body less the feet and head-neck-trunk in individuals of different morphology are argued. This is followed by reviewing the methods used in this study to estimate MOI values of the whole body less the feet and head-neck-trunk segments. Finally, the chapter ends with the thesis' specific objectives.

2.1 Review of body and its segments' MOI estimation methods

The evaluation of segment inertial parameters (e.g., moment of inertia) can be classified into those conducted on cadavers and those in which living subjects were participated. This section focuses mainly on the most common methods that are clinically applied for measuring or estimating of the whole body and its segments' MOI values. The advantage and disadvantages of these methods are also discussed.

2.1.1 Cadaver studies

Attempts to provide data on segments' MOI values began in the 19th century with the onset of cadaver studies (Reid and Jensen, 1990). These studies consisted of sectioning cadavers into segments and measuring the inertial parameters directly. The earliest efforts at this procedure date to the works of Harless (1860), Braune and Fischer (1889), and Fischer (1906), but the most significant development was the work done by Dempster (1955). Dempster (1955) using eight male cadavers conducted the most extensive study on segment inertial parameters to that date. First, he used the method of Reuleaux (1876) to determine the average center of rotation at each joint by fixing two points on a segment and tracking them in two different positions of that segment. Body segments were defined by estimations of the joint centers of rotation. The lengths, masses, and volumes of cadavers' segments were measured. Dempster then calculated the location of the center of mass of the segments using a balance-plate, and the moment of inertia using the compound pendulum technique. Finally, Dempster (1955) created tables reporting the segmental masses as proportions of the total body mass and the locations of the centers of mass and lengths of the radii of gyration as proportions of the segment's lengths. Later, Barter (1957) working with Dempster's data, performed stepwise regression analysis to derive regression equations that more accurately compute segment masses.

Dempster's study is regarded as one of the most comprehensive of the cadaver studies and the proportions reported for the segments inertial parameters have been used extensively in biomechanics research. However, these data were obtained from a small number of old male individuals, all of whom were thin to some extent. Therefore, these proportions might provide substantial errors while applied for populations of distinct morphology from the thin old male cadavers.

Dempster's (1955) method gives mass, COM location, and MOI of the body segments only about a transverse axis. Its application is limited to two-dimensional (2-D) analysis of human movement and cannot be used in this thesis where three principal MOI values are required. However, this method could be applied to provide information in the body segments' mass and COM location.

Many other cadaver studies were conducted since Dempster's work in order to compensate its limitations. The investigations done by Clauser et al. (1969) and Chandler et al. (1979) are noteworthy because they defined body segments using palpable bony landmarks instead of estimated joint centers of rotation. Clauser et al. (1969) dissected a sample of thirteen preserved male cadavers, which permitted sampling over an extended time period and thus a larger sample. The density of the preservation solution was 1.061 $g.cm^{-3}$, which was close to the average density of healthy young men (1.063 $g.cm^{-3}$) as reported by Behnke (1961). Thus, the effect of the preservation solution on total body density was considered to be negligible. However, this assumption may not hold for an obese population. Clauser et al. (1969) measured the mass and COM of each segment using techniques similar to those of the previous studies. Unfortunately, Clauser et al. (1969) did not measure the segment MOI values. Therefore, this method could not be employed for 3-D analysis of human performance, as well.

Chandler et al. (1975) dissected six male embalmed cadavers using segmentation planes similar to those reported by Clauser et al. (1969). They produced the most comprehensive cadaver study of segment MOI values. Segment mass, COM, and anthropometric parameters were also measured. Although

Chandler et al. (1975) cautioned that the data did not reflect the general population due to the limited number of specimens. Vaughan et al. (1992) enhanced the methods used by Chandler et al. (1975) for 3-D kinematic and kinetic investigation of the lower extremity during gait. They developed regression equations to estimate the masses of the lower extremities that included various anthropometric measures (e.g., calf and mid-thigh circumference) in addition to segment length and body mass. Therefore, these regression equations seem to take into account the segment shapes which lead to more personalized inertial properties in subjects with different morphologies (e.g., obese). Hinrichs (1985) used the anthropometric measures from Chandler et al. (1975) as predictor variables for extending the transverse and longitudinal principal MOI values. The computed multiple linear regressions, however, were restricted to two predictor variables because of the small number of specimens. Thus, these regression equations could lead to significant errors while applied for subjects with different body types from the old male cadavers and were not employed in this study.

In general, cadaver studies have the advantage of direct measurements of segment inertial parameters. These measurements can then be used to check the accuracy of the parameters estimates determined from the other techniques while applied to subjects with similar morphology to the cadavers (e.g., old male individuals). The primary disadvantages of the cadaver studies are due to the sample size and the adequacy of the measurements. Samples are small and not representative of the population under investigation. This is particularly so for females, children, adolescents, younger adults, subjects with abnormalities like scoliosis, and population other than Caucasian. For instance, Dempster (1955) dissected eight males cadavers aged 52–83 years, all of whom were emaciated to some extent. The body dimensions and mass distribution vary between different races (Reid and Jensen, 1990), males and females, and period of growth in children (Jensen, 1986). In addition, Pearsall et al. (1996) reported the tissue composition and morphology after death change. Therefore, the regression equations based on cadaver studies would lead to substantial errors while applied to estimate MOI values of living subjects (Yeadon and Morlock, 1989). Their application should be restricted to a similar population from whom the data were obtained. The shortcomings in using cadaver-based methods to estimate MOI values led to alternative techniques where measures were taken from living subjects.

2.1.2 Geometrical modeling

Geometrical modeling techniques are based on the representation of a segment or its components by standard geometric shapes of known density. It is assumed that each segment is a single homogeneous solid. Geometrical modeling of the inertial properties of human body segments was pioneered by the work of Whitsett (1963). He refined an earlier model by Simmons and Gardner (1960) into a 14-segment collection of frustums of right circular cones, elliptical cylinders, spheres, and ellipsoids with inertia parameters calculated for each geometric shape. Segment densities were taken from Dempster (1955). This method appeared to provide more accurate inertial properties compared to cadaver-studies due to accounting for the segments geometry and dimensions.

Later, Hanavan (1964) developed a personalized geometrical model which consisted of 15 geometric shapes dependent on the anthropometry of an individual with segments' mass predicted using Barter's regression equations (1957). Hanavan's model is the most popular geometrical technique (Robertson et al., 2004). By taking additional anthropometric measures, such as malleolus height, knee diameter, mid-thigh circumference, and biacromial breadth, Hanavan (1964) developed equations to compute the three principal moments of inertia. This method is non-invasive, easy, and fast, and appeared to provide reasonable inertial properties of the body and its segments. Mass, COM, and MOI values of the segments about the three principal axes are calculated based on only 25 anthropometric measurements obtained from simple and inexpensive tools. However, the single homogeneous solid assumption fails to take into consideration the shape fluctuations throughout the length of some segments like in thin or obese subjects. Nonetheless, this model can affect MOI estimations of segments with complex shapes because of extending the segment dimensions at its end points to the whole segment's length (Rao et al., 2006).

Hanavan's model (1964) has been enhanced to include more segments and their shape fluctuations. For example, Hatze (1980) developed a 17-segment model based on 242 direct anthropometric measurements and reported its use on a 12-year-old boy and three adults including a female as illustrated in Figure 2.1. The shoulders were treated as separate segments to account for their asymmetry and density fluctuations, and variations in tissue density within segments were considered based on profiles reported by Dempster (1955) and skinfold measures. These permitted Hatze (1980) to make a more accurate assessment of the principal moments of inertia and to account for changes in body morphology due to obesity, pregnancy, and other abnormal states.

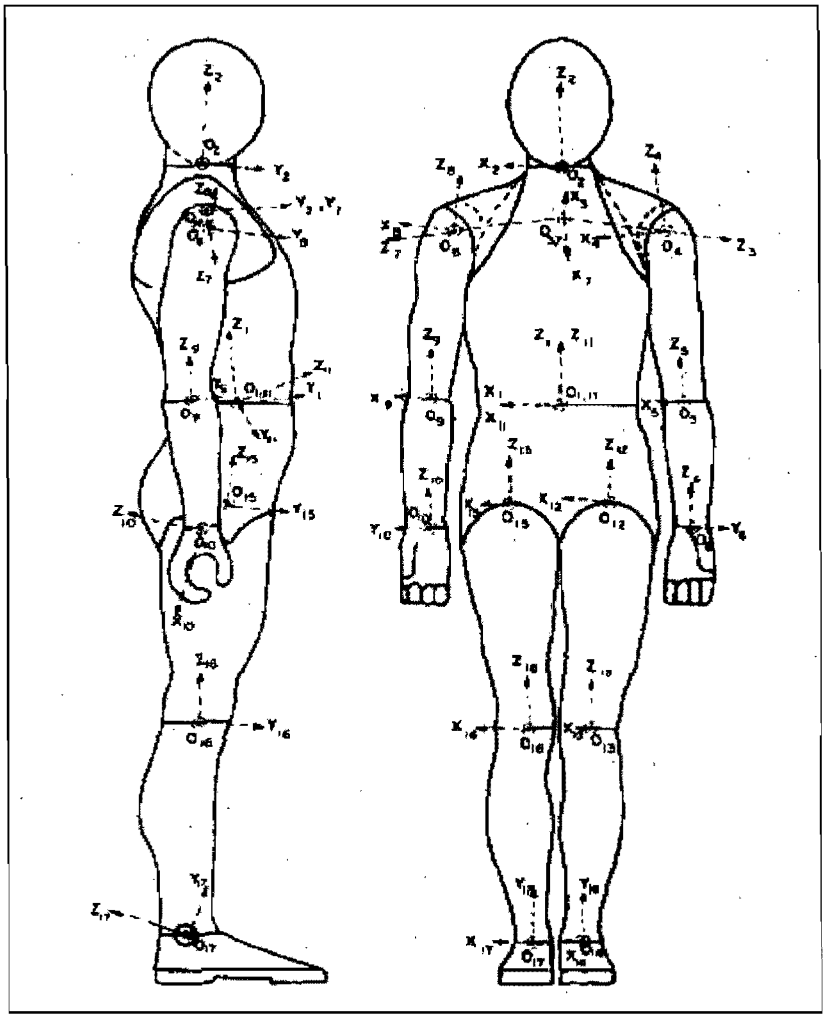

Figure 2.1. Lateral and frontal views of 17-segment of geometrical model (Adapted from Hatze, 1980). The shapes of the segments and local (segment fixed) coordinate systems are also shown.

Hatze (1980) reported an overall accuracy of about 3% and a maximum error of about 5%, only occasionally reaching 11%, for each of the 17 segments' mass, COM position, and MOI values. However, this method is time consuming due to complex data collection and cannot be applied as a clinical approach to estimate the MOI values of the body and its segments.

In summary, geometrical modeling of human segments has the advantage that they can be used for any population and accounts for body segments' shapes. The only assumption to be made is the segment uniform density distribution (Hanavan, 1964). However, errors can be introduced by oversimplification of segment shapes. Hatze, (1980) method requires complex data collection, additional equipment, and extensive anthropometric measurement. Thus, these methods can be applied as a validation technique rather than a clinical method to assess the body and its segments' MOI values.

2.1.3 Imaging techniques

Imaging techniques could be divided into photogrammetry and video-based approaches (radiography and magnetic resonance techniques are discussed under section 2.1.4). The photogrammetric model is based on the assumptions that the body is composed of elliptical zones and the segment densities are known. The elliptical zone approach was used originally by Weinbach (1938) who constructed body profiles, such as for volume and MOI values. Jensen (1978) developed a photogrammetric method in which the human body was composed of 16 segments.

Segments were sectioned into elliptical discs of 20 mm width as showed in Figure 2.2. He used the major and miner axes measured from projected orthogonal photographic images of the body to calculate the inertial properties of the zone. Through the summation of zones and segments, the mass, COM location, and principal MOI values of the segments and body can be estimated. This method takes into account the differences of body segments' volume and shape in individuals with different body morphology (e.g., lean, obese) but still makes the assumption that the segment densities are known. For instance, Jensen (1978) applied segment densities reported by Dempster (1955).

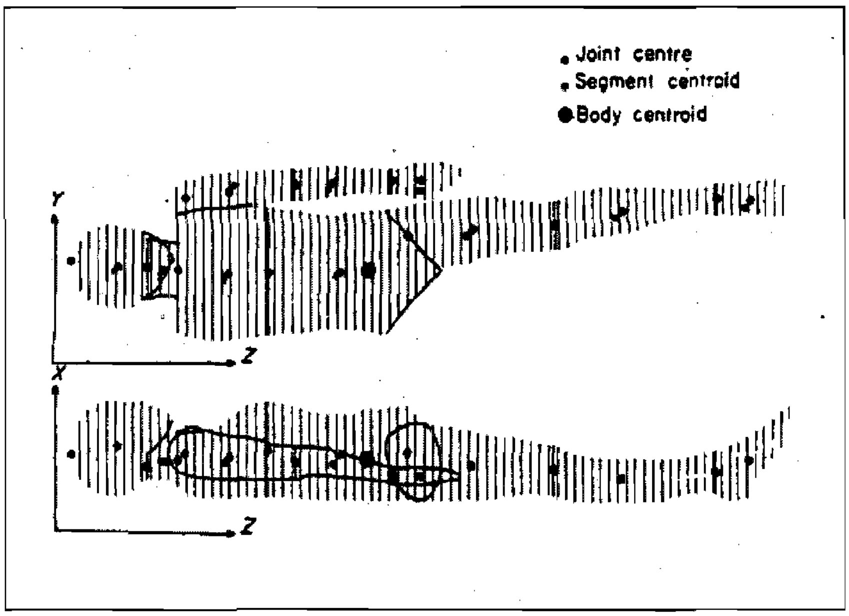

Figure 2.2. Frontal and lateral views of 16-segment photogrammetry model (Adapted from Jensen, 1978). X, Y, and Z indicate antero-posterior, medio-lateral, and longitudinal directions of the body.

Jensen (1988) compared segment masses and MOI values estimated from the photogrammetry to those obtained from regression equations of Morlock and Yeadon (1986) and anthropometric parameters reported by Hanavan (1964). With the exception of the head and feet, all parameters were similar to those of the previous studies. These results suggest that extrapolation beyond the sample age range, 4–20 years, of Jensen's study (1978) should be possible. Finch (1985) applied the photogrammetric method to 15 females of endomorph, mesomorph, and ectomorph to estimate their inertial properties. The predicted values were compared to the results for the adult female from Hatze's study (1980) and the values reported by Plagenhoef (1971) and Zatsiorsky and Seluyanov (1983) and found to be reasonable. Significant differences in inertial properties were found between the different body types by the photogrammetry. This may indicate the capability of photogrammertic method to estimate segment inertial parameters.

The accuracy of the photogrammetric method has also been tested with cadaver and living subject studies. For instance, Tupling et al. (1984) reported similar results of mass estimations based on immersion technique and COM and MOI estimations based on geometrical modeling, compared to photogrammetry. Nonetheless, further investigation is needed to determine the effects of variations in segments' density. Furthermore, the photogrammetric methods were found to overestimate body segments' volume (Kaleps et al., 1984) and the principal MOI values (Hatze, 1980). Generally, this method accounts for the shape fluctuations and estimates the segment inertial parameters comparable with the previous studies. Since this method is time-consuming, it is preferred to be applied as a validation technique instead of a routine method to assess the inertia properties of the segments.

A video-based system for the determination of inertial body segment parameters was presented by Hatze and Baca (1992). They obtained a specific set of anthropometric dimensions from video images and used them as input for the 17-segment model of Hatze (1980). This model accounts for exomorphic and tissue density differences that exist between male and female subjects, for all segmental shape fluctuations and for asymmetries occurring in the geometries of the segments. However, human factors like segment boundaries identification and color thresholds selection were reported as the largest errors in the estimation of the anthropometric dimensions and of the inertial properties using video images (Sarfaty and Ladin, 1993).

Later, Baca (1996) reported that a substantial reduction of these errors could be achieved if appropriate algorithms are applied when processing the images. These algorithms reduce errors originating from optical distortions, inaccurate edge-detection procedures, and user-specific upper and lower segment boundaries or threshold levels for the edge-detection. Baca (1996) developed a video-based technique that can determine the anthropometric dimensions for estimating human body segment parameters precisely. High precision was achieved by finding the location of segment boundaries with sub-pixel accuracy, the implementation of his calibration algorithms, and by taking into account the varying distances of the body segments from the recording camera. Four different views of the subject have to be recorded by a video camera against a black background. These are the anterior view, two lateral views (left and right), and coronal view as illustrated in Figure 2.3. This technique allows automatic segment boundary identification from the video image, if the boundaries are marked on the subject by black ribbons. These anthropometric dimensions are then used as input for the Hatze's model (1980) to compute body segment parameters (volumes, masses, the three principal MOI values, and the 3-D local coordinates of segmental COM).

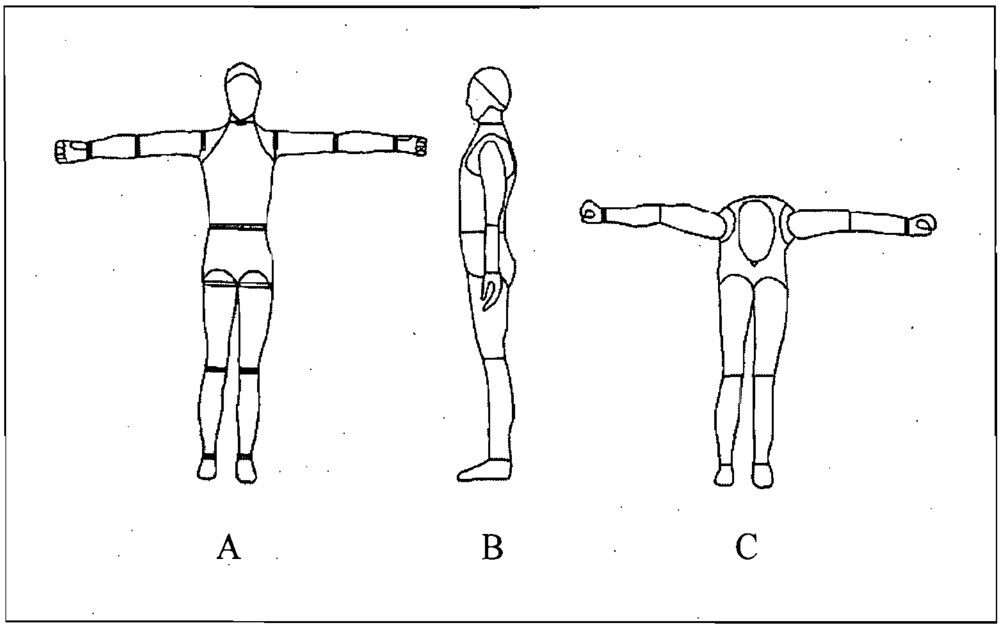

Figure 2.3. Recording positions, A) anterior view, B) lateral view (left side), C) coronal view. (Adapted from Baca, 1996)

The method reported by Baca (1996) was found to provide relatively accurate anthropometric dimensions from video images where these data can be used to estimate individualized segment inertial properties. The anthropometric values computed by this method do not differ much from those based on direct anthropometric measurements. However, the segment inertial parameters errors obtained from video-based system were considerable compared to direct anthropometric measures. For instance, mass, COM position, and the MOI values had maximum errors of 7.9, 8.0, and 13.7%, respectively, compared to Hatze's model where direct anthropometric measurements were obtained. In addition, application of this method accompanied by Hatze's model (1980) still needs 220 anthropometric measures of subject. This is a very time-consuming technique that cannot be used as a clinical method to estimate segment inertial parameters. More practical techniques that need less acquisition time and have the ability to estimate segment inertial parameters accurately are required.

2.1.4 Scanning techniques

Though radiation techniques are mainly invasive, they can be used to estimate inertial properties of body segments. Radiation techniques are divided into: gamma scanning method (Zatsiorsky and Seluyanov, 1983, 1985), dual energy x-ray absorptiometry (Durkin et al., 2002; Durkin and Dowling, 2003), biplanar radiography (Dumas et al., 2005), MRI (Martin et al., 1989; Mungiole and Martin, 1990), and CT imaging (Pearsall et al., 1996). In all these techniques some kinds of radiations (e.g., x-ray or gamma-ray) pass through the subject's body. These techniques are based on the assumption that the ability of an object to absorb or attenuate high energy radiated rays is proportional to the density of the object and relatively independent of its composition.

Casper et al. (1971) were the first to estimate the mass, COM position, and MOI values of an object by gamma ray imaging. Inertial parameters for wood, metal, and plastic objects were within ±1% of criterion values. To determine the ability of gamma mass scanning in living tissue, Brooks and Jacobs (1975) applied this technique to calculate inertial properties of a leg of lamb. They compared scanner estimates of mass, COM position, and MOI values with direct measurements of the segment and found errors of 1, 2.1, and 4.8%, respectively. Later on, gamma mass scanning technique was applied by Zatsiorsky and Seluyanov (1983) for tests on humans. They presented data from an extensive study on the 3-D body segment parameters of 100 males and 15 females college-aged Caucasian. The sample had a mean age of 23.8 years (SD = 6.2). To compute mass distribution, gamma mass scanning was used to quantify the density of incremental slices of each segment. Zatsiorsky and Seluyanov (1983) used multiple linear regression models, with weight and stature as predictor variables, to estimate the mass, COM, and three principal MOI values for a total of 16 segments. These regressions equations were supplemented by a further set in which segment-specific anthropometric measures were used as predictor variables (Zatsiorsky and Seluyanov, 1985). Comparisons of the inertial properties estimations of Zatsiorsky and Seluyanov (1983, 1985) with cadavers (e.g., Dempster, 1955; Clauser et al., 1969) and living subjects (Bernstein et al., 1931; Plagenhoef, 1971) studies indicated that the results for most parameters were within the range of the other methods.

The inertial parameters reported by Zatsiorsky and Seluyanov (1983, 1985) were obtained from a large sample of young adult Caucasians population. Thus, their application to a similar population would provide reasonable values. However, they could be misleading to individuals with different body morphology. This technique requires expensive instrumentation and specialized operators, and may involve high radiation levels. Thus, this method can be applied as a validation technique rather than a routine method to assess the segments inertial parameters.

Dual energy x-ray absorptiometry (DEXA) was another radiation approach developed to calculate inertial properties of the body segments (Durkin et al., 2002; Durkin and Dowling, 2003). The whole body DEXA scan is performed with the subject lying supine on the scanner table with palms facing the table as shown in Figure 2.4. Durkin and Dowling (2003) reported that this technique has the ability of measuring inertial properties with great accuracy, while the radiation exposure for a whole body scan is 1/10 of a chest x-ray. Though, regression equations were provided to estimate inertial properties of various populations, data were not taken on obese and subjects with structural abnormalities. This method needs expensive tools and provides the MOI values of segments only in frontal plane. Therefore, this provides only 2-D measurements of the segment inertial parameters.

To compensate the limitation of DEXA related to 2-D measurements of inertial properties, a biplanar radiography method was proposed by Dumas et al., (2005). Simultaneous low-dose frontal and sagittal radiographs were obtained with EOS^{TM} system from thigh segment of young males and females. The 3-D inertial parameters computed from biplanar radiographic were consistent with those of gamma mass scanning and DEXA. However, this method still needs expensive instruments and high skill operators.

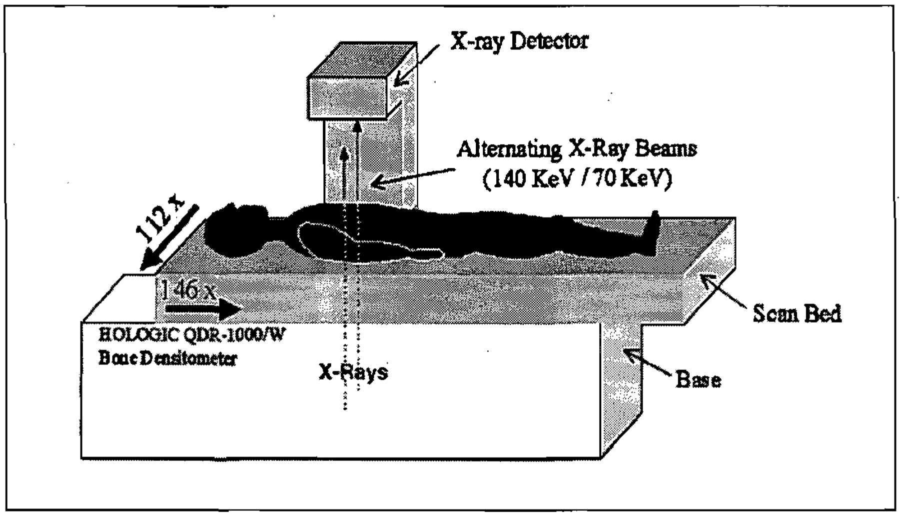

Figure 2.4. Example of a whole body scan of human using a dual energy x-ray absorptiometer. Attenuation coefficients based on x-ray absorption values are recorded in elements of x-ray detector, resulting in samples of mass per whole body scan. (Adapted from Durkin et al. 2002)

MRI and CT images have also been used on living subjects. Mungiole and Martin (1990) determined the inertial properties of lower leg of 12 males by means of MRI and concluded that the values were within the range reported for the cadaver studies. Reid (1984) and Pearsall et al. (1994, 1996) also reported MRI and CT techniques as favorable approaches for estimation of inertial parameters of the trunk in males and females. Transverse CT images were collected at 1 cm intervals, while in MRI transverse slices of 10 mm were acquired. Cheng et al. (2000) developed a MRI technique, where magnetic resonance images were scanned at 20 mm intervals, for measuring segment inertial parameters of Chinese male subjects. They compared their calculations with those of previous studies (Dempster, 1955; Clauser et al., 1969; Martin et al., 1989; Pearsall et al., 1994) and found larger mass percentages for upper arm (4.0%) and thigh (13.6%), and smaller MOI values for the shank.

These approaches seem to be appropriate to measure segment inertial parameters in populations with different morphology. The accuracy and precision of CT and MRI were evaluated by Zhu et al. (1986). They reported smaller errors for CT and the scan time is less. However, CT has health risks of radiation exposure. The cost and availability of facilities for these techniques restrict their use in clinical situations. These methods could be applied to validate the other methods.

In summary, the radiation techniques have the ability to provide accurate 3-D measures of the inertial parameters with the exception of DEXA. They can be applied in different populations. However, these methods require expensive tools and have health risks due to radiation exposure except for MRI. Though, these methods provide insight into the accuracy of the other estimation methods, their application in a clinical set-up could be difficult. Developing personalized clinical methods such as oscillation techniques to calculate segment inertial properties could be considered as a way to overcome the drawbacks of the previous studies.

2.1.5 Oscillation techniques

Segment moments of inertia have also been estimated using the quick-release method (Drillis et al., 1964) and oscillation technique (Hatze, 1975). In both techniques a body segment oscillates while the muscles are relaxed, so that they do not influence the acceleration of the oscillated limb and damped oscillations. Recently, Monnet et al. (2007) developed an identification method to estimate the 3-D MOI values of the upper limb during flexion/extension, abduction/adduction, and circumduction. This method consists in solving a redundant system by numerical computations. In these three methods a segment should be able to oscillate freely. Thus, they can be used only to measure MOI values of extremities because applying them to other body parts would be difficult.

Bouisset and Pertuzon (1968) restrained the forearm and hand segment by using a fixed moment of force that was suddenly released. The MOI value about transverse elbow axis was then calculated from the peak angular acceleration and corresponding moment ($I = M/\alpha$). This technique assumes that the oscillating limb is affected only by its inertial properties. Using peak angular acceleration may provide inaccurate MOI values of the segment owing to the effect of other factors (e.g., passive muscle stiffness) rather than the segment's inertial properties. In addition, peak angular acceleration has the highest level of noise that could cause significant errors on the MOI estimations. Hatze (1975) estimated MOI values of lower extremity and leg and foot segments from damped oscillations. The results of this study were comparable with those of cadavers (Dempster, 1955) and Hanavan's model (1964). This technique requires that a body part be set into oscillation with an instrumented spring. The equations based on small oscillation theory are then used to estimate the properties of the segment. Later, Alum and Young (1976) developed the oscillation technique by using forced sinusoidal oscillation to determine the MOI value of the forearm and hand about the elbow axis of four subjects. This technique was also further developed by Peyton (1986), where it involved coupling the limb segment to an elastic mechanical device. The resulting system has a lightly damped oscillatory response from which its resonant frequency can be measured and used to determine the MOI of the limb. The MOI results of these studies were favorably comparable with those of Dempster (1955) and Bouisset and Pertuzon (1968). However, these methods cannot be applied for the trunk segment and provide the MOI values about the transverse axis that could be applied for 2-D analysis of the extremities.

Several oscillation techniques have also been developed to calculate the whole body moment of inertia. For instance, Matsuo et al. (1995) measured the MOI values of whole body directly using an oscillation table (Figure 2.5). The measurement error was less than 0.5%. This technique can be applied clinically in every group of populations to determine the MOI values of the whole body about sagittal and frontal axes only as illustrated in Figure 2.6. Since, the body should be fixed during the trials, calculating the moment of inertia in vertical axis would be a difficult task. Though this method measures the MOI values with great accuracy, the oscillation table is expensive, limited to the whole body, and the sagittal and frontal axes.

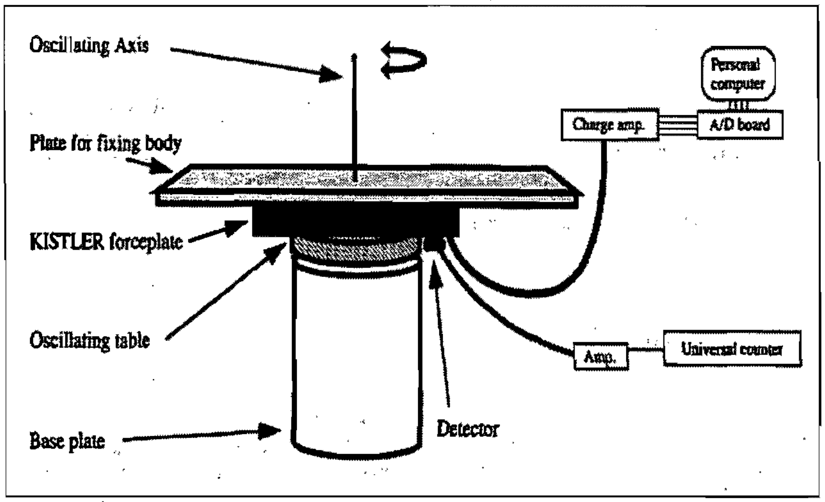

Figure 2.5. Illustration of the measurement device to measure moments of inertia of the whole body (Adapted from Matsuo et al., 1995)

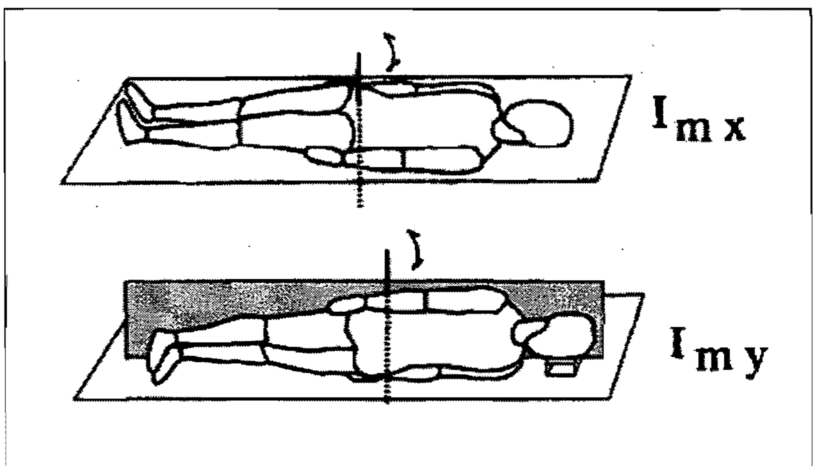

Figure 2.6. The measurement postures of moment of inertia (Adapted from Matsuo et al., 1995)

Another approach to calculate moment of inertia of the body and its segments is pendulum technique. This method is able to calculate the MOI values of the body about the antero-posterior and medio-lateral axes through its COM. Plagenhoef (1971) computed the MOI values of a rigid body using direct pendulum method by measuring the period of oscillation. This method is simple and calculates the MOI values of a rigid body accurately. Direct pendulum technique has been used to calculate the segments' MOI values in cadavers (Dempster, 1955; Chandler et al., 1975). This method could also be applied to calculate the MOI values of the body in in-vivo. For instance, Smith (1957) used pendulum method to determine the MOI of human body about the ankle. Direct determination of the MOI of the living body about the ankle axis by this method would require suspension of the subject upside down. Use of the parallel axis theorem makes this procedure unnecessary as shown in Figure 2.7. This method cannot be used to calculate the MOI values of the body about the longitudinal axis through its COM since the distance between body COM position and the axis of rotation cannot be calculated.

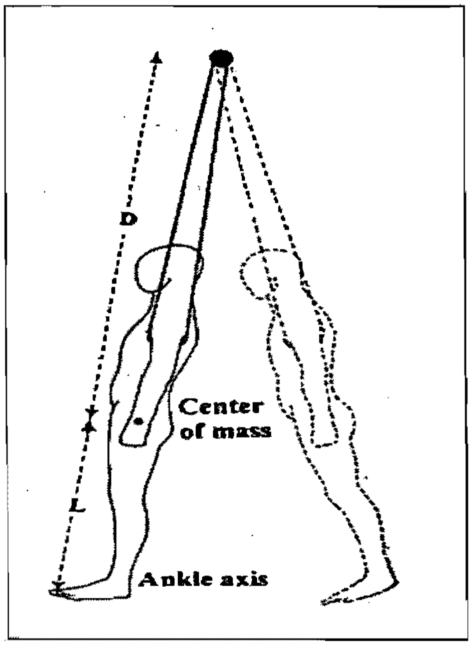

Figure 2.7. Determination of the moment of inertia of the human body by the pendulum method (Adapted from Zatsiorsky, 2002)

Two new personalized in-vivo methods based on 3-D inverse pendulum model were developed in this study for the clinical assessment of the whole body MOI values. These methods make use of the inverse dynamics and angular momentum equations to estimate the MOI values. These are both in-vivo and personalized so that they can be applied to populations with different body morphologies. The inverse dynamics method is based on 3-D inverse pendulum model of Barbier et al. (2003) in which the whole body less the feet COM oscillates about the ankles. In Barbier et al. (2003) the body is divided into two segments; the whole body less the feet and the feet. Then, the 3-D excursion of the COM of the body less the feet is estimated from external forces, ankle muscle moments, and inertial properties of the segments using Euler's equation of motion.

Conversely, if the COM angular accelerations are determined from videography by this approach, then the moments of inertia of the oscillating whole body can be estimated for each individual.

The second method applies the angular momentum equation. Again the body is considered as two segments of the whole body less the feet and the feet. It consists of tracking the body and measuring the moments during self-imposed oscillations about the ankles by means of a video-based system and force-plate. Then, the moments of inertia of the body is calculated from video data and the integration of the moments obtained by the force-plate. Since the feet are fixed to the ground during the oscillations about the three principal axes of rotation, these two methods can be applied to estimate the whole body MOI values but the feet. Both new methods can be used to estimate the MOI values of the body for any kind of populations. Once the whole body MOI values are known, body segments' MOI can be obtained by applying the parallel axis theorem.

In summary, the regression equations based on cadaver studies would lead to substantial errors while applied to estimate MOI values of living subjects (Yeadon and Morlock, 1989). Their application should be restricted to a similar population from whom the data were obtained. Geometrical modeling of human segments can be used for any population and accounts for the body segments' shapes. The only assumption to be made is on segment uniform density distribution (Hanavan, 1964). These methods require complex data collection and additional equipment (Hatze, 1980). Thus, geometrical models can be applied as a validation technique rather than a clinical method to assess the moments of inertia of body segments. Photogrammetric method (Jensen, 1978) accounts for segment shape fluctuations, an advantage in segments with complex contour (e.g., trunk). However, this method was found to overestimate body segments' volume (Kaleps et al., 1984) and the principal MOI values (Hatze, 1980). Radiation techniques have the ability to estimate segment inertial parameters with reasonable accuracy. However, these methods have health risks of radiation exposure except for MRI. Radiation techniques require high skill operators and are expensive. Oscillation techniques have the capability of calculating MOI values of the segments. These techniques, however, cannot be used for the trunk segment or are limited to the AP and ML axes. Inverse dynamics and angular momentum methods can provide in-vivo and personalized information on the body less the feet and its segments MOI values in populations with different body types.

2.2 MOI values of the whole body less the feet and head-neck-trunk in individuals of different morphology

The MOI values of a body segment depend on its mass and on the distribution of mass within the body with respect to the axis of rotation. Durkin and Dowling (2003) reported that segments' mass expressed as a percentage of the total body mass, and radii of gyration of body segments are different in people of various body morphologies. The MOI values vary within and between different populations, such as males and females, different races, and sports groups (Reid and Jensen, 1990). According to Zatsiorsky (2002) the whole body MOI values are approximately proportional to the subject's height to the third and fifth power, respectively. Consequently small changes in body size could result in large changes in the MOI values.

BMI is often used to classify individuals into lean, normal and obese types. Current MOI methods have been developed mostly based on a specific sample such as lean (e.g., Dempster, 1955) or normal (Zatsiorsky and Seluyanov, 1983) BMI ranges. Therefore, these methods produce substantial inaccuracies when applied to people with different morphologies than those of the original studies (Ganley and Powers, 2004). The inverse dynamics and angular momentum methods could be applied to estimate the MOI values of subjects with different body morphology (i.e., lean and obese). Therefore, to evaluate the effect of methods on MOI values of the whole body less the feet and head-neck-trunk, they can be tested in distinctive body types. These effects could provide insight into the sensitivity of the MOI models applied to population of different morphology.

In literature, the most variability in the body segment inertial parameters has been reported for the trunk segment (Zatsiorsky, 2002). This could be due to difficulty in modeling the inertial properties of the trunk. There are several reasons for this as described here. Due to the lungs, the density of the upper parts of the trunk is much lower than the density of the middle and lower parts (Pearsall et al., 1996). The inertial properties of the trunk vary with the inspiration and expiration. The internal organs are displaced within the trunk when the body changes its orientation in space. The trunk is difficult to separate from other parts of the body and can barely be considered as a single rigid body. Therefore, any assumption made by the modeling methods on the segment's density profile could provide inaccuracies in the MOI estimations. Though it includes a considerable mass portion of the body, the relative trunk mass was reported from 42.2% (Pearsall et al., 1994) to 52.4% (Chandler et al., 1975). Even in a homogenous sample of adult men, the relative trunk mass was ranged from 35.8 to 48.0% (Pearsall et al., 1994). Thus, it can strongly be expected that the MOI values of the trunk are affected by the used methods and vary substantially in population with various morphology. These methods were those of de Leva (1996), Hanavan (1964), Jensen (1978), and the angular momentum method.

2.3 Methods to estimate MOI values of the whole body less the feet and head-neck-trunk

To determine the accuracy of the MOI values obtained from the inverse dynamics and angular momentum methods, their results should be compared to those of the other techniques. In the absence of criterion measures for the MOI values, the question of the accuracy of the estimates has to be approached through a variety of validation procedures. The most common of these involves checking the estimates of whole body MOI values against the more readily obtainable whole body measures. This could serve as a means of gaining insight into the accuracy of the existing models.

In selecting a method or set of prediction equations for inertial parameters, the age and sex of the sample should be primary consideration. The use of living subjects offers the possibility of sampling populations more adequately by increasing the size of the sample and purposive selection. Some techniques are potentially hazardous (e.g., CT imaging, gamma-ray scanning, DEXA) or their application is restricted (e.g., MRI). Thus, in the present study three well-known methods based on living subjects were employed to verify the accuracy of the MOI values of the body less the feet and head-neck-trunk segments obtained from the novel methods. These methods involved anthropometric method of de Leva (1996), geometrical model of Hanavan (1964), and the photograpmmetric method of Jensen (1978).

Segment MOI values in young males (n = 100) and females (n = 15) were estimated using gamma-ray scanning technique (Zatsiorsky and Seluyanov, 1983, 1985). These data are the only available and comprehensive set of 3-D inertial parameters regarding young adult Caucasian. Predictions from the radiation studies by Zatsiorsky and Seluyanov (1983, 1985) appear to be the most appropriate for adults (Reid and Jensen, 1990). The anthropometry of their sample is similar to the results of surveys conducted in other countries (Reid and Jensen, 1990) and those of the normal BMI range subjects of this study. Later, de Leva (1996) adjusted the segments endpoints of the Zatsiorsky et al., (1990a) so that they corresponded to joint centers of rotation. Therefore, the de Leva method was chosen (as a criterion) to test the accuracy and similarity of the other techniques for assessing the MOI values of the whole body less the feet and head-neck-trunk segments.

Geometrical models are able to calculate segments' volume and take into consideration their differences in subjects with distinct body types. A simple but the most applied geometric model was developed by Hanavan (1964). He applied standard forms to represent body segments to estimate their 3-D moments of inertia. Hanavan's model (1964) includes 25 anthropometric measures of the subject's body and can provide reasonable estimations of the body segments' moments of inertia. This method takes into consideration the segments geometry of subjects with different body types (e.g., lean and obese). Thus, it was used in this study to estimate the whole body less the feet and head-neck-trunk MOI values. To account for fluctuations in segments' shape and especially those with complex contour like trunk, a photogrammetric technique (Jensen, 1978) was also applied. This technique was shown to provide reliable estimates of inertial parameters in children (Jensen, 1989).

To calculate the head-neck-trunk MOI values, its mass and COM location are required. It was reported that mass and COM position of the trunk vary based on different methods (Pearsall et al., 1994; Zatsiorsky, 2002). Thus, the effect of the methods to estimate the segment's mass and COM position along their longitudinal axis in different morphological groups needs to be first verified. These methods were the anthropometric methods (de Leva, 1996; Dempster, 1955), the geometric model (Hanavan, 1964), the photogrammetric method (Jensen, 1978), and a new force-plate technique. A new force-plate technique was developed to avoid regression equations and prediction methods to estimate the body segments' mass and COM location.

2.4 Specific objectives of this thesis

Two novel techniques for estimating MOI values of the whole body less the feet and head-neck-trunk segments are the central interest of this thesis. It is hypothesized that the inverse dynamics and angular momentum approaches are sensitive enough to provide in-vivo personalized estimates of the MOI values in population with different morphology. To avoid estimating methods, the segment mass and COM location are measured using a new force-plate technique.

The first objective of this study is to test the effect of the modeling methods to estimate the MOI values of the whole body less the feet obtained from individuals of different BMI representing normal, lean, and obese morphological types. These methods are those of de Leva (1996), Hanavan (1964), Jensen (1978), and two new methods based on an inverse pendulum approach.

The second objective was to verify if the modeling methods behaved similarly to the de Leva (1996) method (as a criterion) by means of Pearson coefficient of correlation for each morphologic group.

The accuracy of the predicting methods to estimate the MOI values of the head-neck-trunk can be evaluated in subjects with distinct morphologies. It is hypothesized that estimating models will have their own effect on the MOI values. Thus, the third objective of this study is to test the effect of the modeling methods to estimate the MOI values of the head-neck-trunk in individuals of different morphological types. These methods are those of de Leva (1996), Hanavan (1964), Jensen (1978), and angular momentum approach. Since to calculate the head-neck-trunk's MOI values, its mass and COM location are required, the effect of methods to estimate the segment's mass and COM position along their longitudinal axis in different morphological groups is first verified. These methods are those of de Leva (1996), Dempster (1955), Hanavan (1964), Jensen (1978), and a new force-plate technique. The behavior of the methods to estimate mass and COM position was compared to the de Leva (1996) method for each morphological group.

The fourth objective is to verify if the methods used to estimate head-neck-trunk's MOI values behaved similarly to the de Leva (1996) method by means of Pearson coefficient of correlation for each morphologic group.

Chapter 3

3. METHODS

This chapter begins with the subjects' anthropometrical characteristics. This is followed by the methods applied to estimate the MOI values of the whole body less the feet. The anthropometric (de Leva, 1996), geometric (Hanavan, 1964), and photogrammetric (Jensen, 1978) methods are first described. Then, two novel methods based on inverse dynamics and angular momentum equations are detailed for self-imposed oscillations. To apply the latter in the calculation of the MOI values of head-neck-trunk, the mass and COM position of this segment are required. A new force-plate technique developed to avoid using anthropometric tables to calculate these values is presented. In summary, five methods to estimate the whole body moment of inertia properties and four methods (de Leva, 1996; Hanavan 1964; Jensen, 1978; and angular momentum) for estimating the MOI values of the head-neck-trunk will be compared in subjects with three types of body morphology.

3.1 Subjects

Twenty-three able-bodied adults consisting of 19 males and 4 females with no previous orthopedic ailment or impairment that could affect their standing posture and the self-imposed oscillations participated to this study. Prior to the experimentation, all procedures of the protocol were explained to each subject and an informed consent approved by the Sainte-Justine Hospital Ethics Committee (No. 2341, as presented in Appendix A) was obtained.

Subjects' height and weight were measured to calculate their body mass index (BMI = mass/height^2) using a height gauge and force-plate, respectively. Then, subjects were divided into three morphological body types. A lean subject had a BMI of 18.5 kg/m^2 or less; a normal value was within a BMI range of 18.5 and 24.9 kg/m^2, while an obese individual had a BMI greater than 30 kg/m^2 (Heyward et al., 2004). Subjects with a BMI ranging from 25 to 29.9 kg/m^2 corresponding to an overweight group were excluded form this study. Obese subjects were preferred over the overweight ones because of their higher MOI values. The female subjects were in the lean group. Table 3.1 provides the anthropometric characteristics of each morphological group. No significant differences were noted on age and height of the subjects between the morphological groups.

Table 3.1. Means and (standard deviations) of age, mass, height and body mass index (BMI) of normal, lean and obese morphological groups along with the number of subjects (*n*) in each morphological group.

Morphological group	Age (yr)	Mass (kg)	Height (cm)	BMI (kg/m^2)
Normal (*n* = 7)	32.4 (7.7)	75.4 (2.8)	177 (3.8)	23.9 (1.1)
Lean (*n* = 8)	29.8 (7.7)	53.2 (8.0)	172 (11.8)	17.8 (1.2)
Obese (*n* = 8)	33.3 (3.6)	100.0 (4.3)	178 (4.9)	31.8 (1.9)

3.2 Moment of inertia of the whole body less the feet

The anthropometric method of de Leva (1996), geometric model of Hanavan (1964), and photogrammetric method of Jensen (1978) were applied to estimate the whole body MOI values less the feet. Since these methods are conventional and well known, a brief description is presented. Two methods based on 3-D inverse pendulum model are proposed here for the first time to estimate the whole body MOI values. These methods are developed from the work of Barbier et al. (2003) where the human body is modeled as a 3-D inverse pendulum representing a point mass oscillating about the ankles.

3.2.1 Anthropometric, geometric, and photogrammetric methods

To estimate the whole body less the feet MOI values, the segment inertial parameters (SIP) of the body segments were calculated by regression equations of de Leva (1996) as follows. The segment lengths of all subjects were measured using a tape meter. The mass and COM position of the segments were calculated respectively as a percent of total body mass and the ratio of the segment length with respect to the proximal end as defined by de Leva (1996). Then, using the radii of gyration with respect to the each segment length, the MOI values along the three axes of rotation at their COM position were estimated. The reader is referred to de Leva (1996) for further information.

Since the moments of inertia of the whole body less the feet are reported with respect to its COM, the 3-D positions of the whole body less the feet and its segments need to be calculated. The vertical COM position (z) of the whole body less the feet with respect to a point lying midway between the ankles was calculated by

$$z = \frac{1}{M} \sum_{i=1}^{n} (m_i z_i) \qquad (1)$$

where M, m_i, and z_i are whole body less the feet mass, individual segment masses, and the segments' COM position along their longitudinal axis, respectively.

Then, the COM horizontal positions of the whole body less the feet and its segments were computed in a global coordinate system. Reflective markers were put over the anatomical landmarks as defined by de Leva (1996) to identify the 3-D coordinates of the joint centers in a global coordinate system by means of video-based system. The origin of this coordinate system was located at the mid-point between the two ankles. Subjects were instructed to stand quietly over the center of force-plate with arms beside the trunk and feet were parallel and 10 cm apart from each other for a period of 20 s. The mean values of the centers of pressure (COP) coordinates were considered as the whole body COM positions along the horizontal directions (Murray et al., 1967). Afterwards, the horizontal COM coordinates of the whole body less the feet were calculated by subtracting the corresponding values of the feet using equation (1). The distance between the body segments' COM and those of whole body less the feet in horizontal directions were then calculated. Finally, the whole body less the feet MOI values about the

COM ($I_{b-f_{/COM}}$) for the three principal axes were calculated using the parallel axis theorem by

$$I_{b-f_{/COM}} = \sum_{i=1}^{n} I_{COM_i} + \sum_{i=1}^{n} m_i r_i^2 \qquad (2)$$

where I_{COM_i}, and r_i are the individual segments' centroidal moments of inertia, and the distance between each segment COM along each axis and the whole body less the feet COM, respectively.

Another model to be applied was that of Hanavan (1964). It consists of 15 geometric shapes as illustrated in Figure 3.1. Twenty-four anthropometric measurements of the subjects were acquired according to the original document (Hanavan, 1964) to estimate the whole body less the feet MOI values. These were ankle circumference, arm circumference, buttock depth, chest breadth, chest depth, elbow circumference, fist circumference, forearm length, knee circumference, head circumference, hip breadth, shoulder height, sitting height, sphyrion height, stature, substernal height, thigh circumference, total body mass, tibiale height, greater trochanter height, upper arm length, waist breadth, waist depth, wrist circumference. These anthropometric measures were applied to calculate the COM positions and the MOI values along the three principal axes of each segment. The segment masses were estimated as a ratio of the whole body mass (Hanavan, 1964). Again the reader is referred to (Hanavan, 1964) for further information.

After estimating of SIP for each segment, the same approach as described for the de Leva (1996) method were followed to estimate the MOI values of the whole body less the feet.

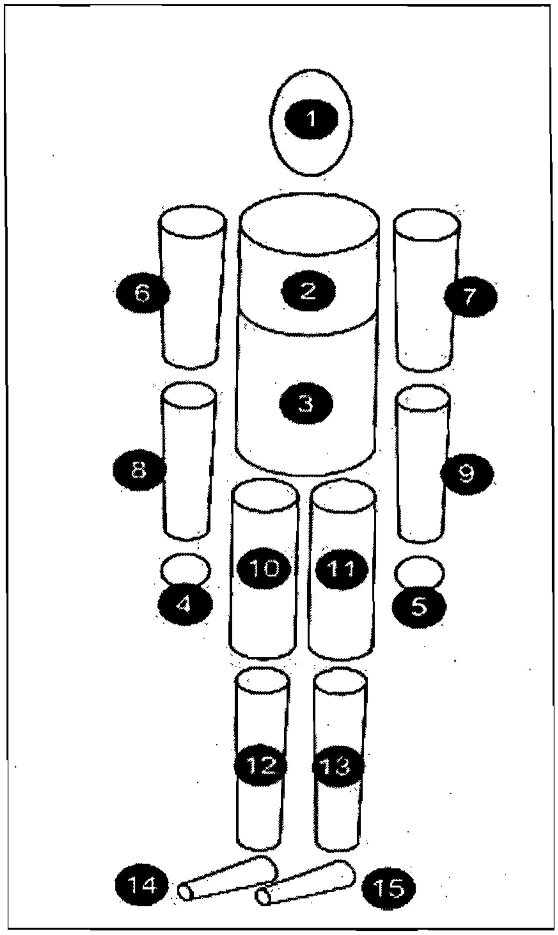

Figure 3.1. Hanavan's (1964) geometrical model of the body. (Adapted from Robertson et al., 2004). The body segments are considered as geometric shapes and numbered as defined by Hanavan (1964).

The photogrammetric method developed by Jensen (1978) was applied to estimate the whole body less the feet MOI. The optical axis of two digital cameras was located perpendicular to the mid-sagittal and mid-frontal planes of the subjects' body, respectively. To remove the distortions at the margins of the images, the cameras-subject distance was approximately 4.5 meters. Thirteen markers of 16 mm diameter were put over the tip of the second foot finger, lateral malleolus, femoral condyle, greater trochanter, the tip of the middle hand finger, ulnar styloid, the greatest projection of the medial humeral epicondyle, humeral head, omphalion, xyphion, suprasternale, chin area aligned to the horizontal plane going through the first cervical spine (C1), and vertex (top of the head). These were applied on the right side of the body to facilitate the segment boundaries identification during images' digitization as illustrated in Figure 3.2. These defined 16 segments, namely the head, neck, upper trunk, lower trunk, upper arms, lower arms, hands, thighs, shanks, and feet. The division of each segment follows the basic procedures recommended by Dempster (1955).

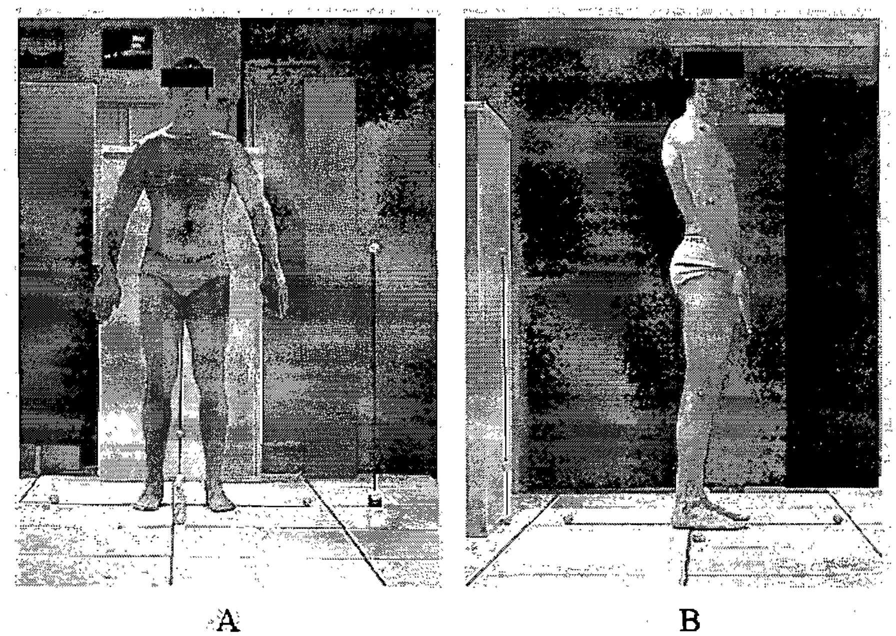

Figure 3.2. A) Frontal, B) Lateral views of a subject for the photographs. The pictures were taken simultaneously by two digital cameras. The cameras-subject distance was 4.5 meters.

Four sticks of 1 meter length were located horizontally on the ground, and vertically beside and behind the subject corresponding approximately to the mid-sagittal and mid-frontal planes of the body. They were used as scales to determine the segments' length. Body segments were digitized by contouring on both photographs using dedicated software (Slicer, Laurentian University, Sudbury, ON, Canada). Once the contours were digitized and scaled, body segments were horizontally sliced every 5 mm by the software. Each slice's volume is obtained using

$$V_{slice} = \pi \cdot r_x \cdot r_y \cdot r_z \qquad (3)$$

where V_{slice}, r_x, r_y, and r_z are slice volume (m^3), the anterior-posterior (AP) radius (m), medio-lateral (ML) radius (m), and half the slice's thickness (2.5×10^{-3}m), respectively. Segment volume is computed by

$$V_{segment} = \sum_{i=1}^{n} V_i^j \qquad (4)$$

where V_i^j is the volume of the i^{th} slice of the j^{th} segment. The slice mass is calculated by

$$m_{slice} = \rho_{segment} \cdot V_{slice} \qquad (5)$$

where $\rho_{segment}$ is the assumed uniform density of the segment based on Dempster (1955). Afterwards, the segment mass is calculated by

$$m_{segment} = \rho_{segment} \cdot V_{segment} \qquad (6)$$

The COM coordinates of each slice are computed in the global coordinate system as follow

$$COM_{slice} = \begin{matrix} x_{s\min} + r_x \\ y_{s\min} + r_y \\ z_s + r_z \end{matrix} \qquad (7)$$

where $x_{s\min}$, and $y_{s\min}$, are the smallest slice's coordinates with respect to an inertial reference frame along AP and ML axes, respectively, and z_s is the altitude of the given slice. Given the slices' COM coordinates, each segment COM location is calculated from equation (1) and for each axis. Then, slice's moments of inertia about the centroidal axes were computed as defined by Jensen (1978). Using the parallel axis theorem (equation 2) the segments MOI values and then whole body MOIs less the feet were calculated. The reader is referred to Jensen (1978) for more detail information.

3.2.2 Inverse dynamics and angular momentum methods

The inverse dynamics and angular momentum methods are based on tracking body segments during self-imposed rotations at the ankles about the antero-posterior (AP), medio-lateral (ML) and longitudinal (LG) axes. These methods are based on that of Barbier et al. (2003) where the human body is modeled as a 3-D inverse pendulum representing a point mass oscillating about the ankles (Morasso et al., 1999; Brenière, 1996). Since the feet are immobile during upright standing, the inverse dynamics and angular momentum methods provide an estimation of the moment of inertia of the whole body less the feet.

In the 3-D inverse pendulum model described by Barbier et al. (2003) the ankle reaction forces and muscle moments are eliminated by substitution, leaving the 3-D COM coordinates as unknowns. This approach can be applied here but by having the moment of inertia as the unknown because the 3-D excursion of the center of mass can be tracked by a video system. Euler's equation of motion ($\Sigma M = I\alpha$) is taken at point C located at the midpoint between the two malleoli as shown in Figure 3.3. The position of point C does not affect the calculations since the moment transfer terms will be taken in the summation of moments of the COM of the body (Barbier et al., 2003). Consequently, setting the origin between the ankles does not exclude conditions involving individuals who usually do not stand with 50% of their weight on each foot.

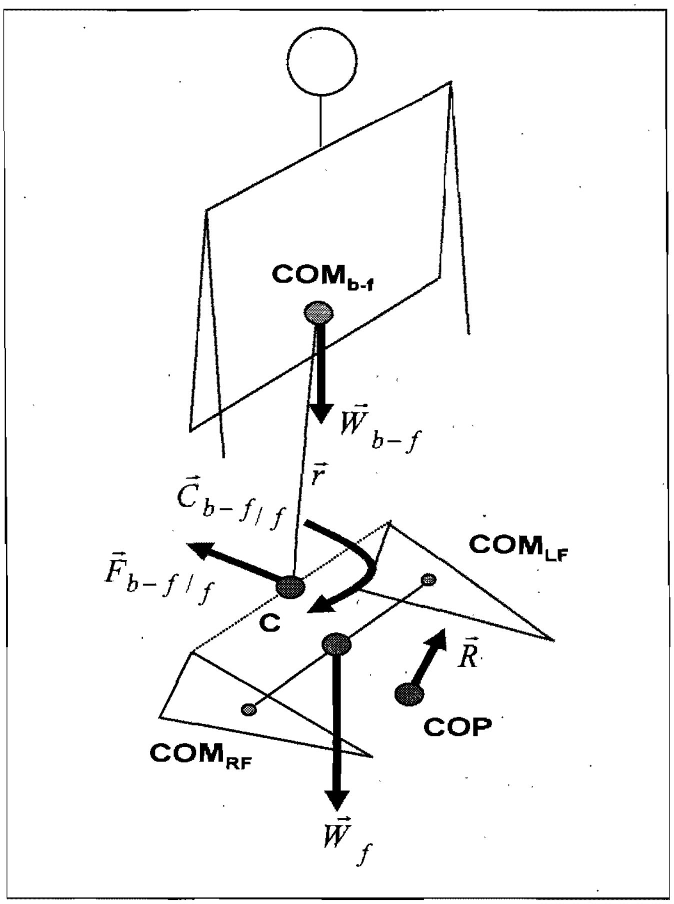

Figure 3.3. Inverse pendulum model of the human body representing the COM oscillations about a point (C) lying midway between the ankles. Symbols are described in the text.

For the feet, the summation of moment at the point C and for each oscillation is

$$\vec{C}_{b-f_{/f}} + \vec{M}_{\vec{R}_{/C}} + \vec{M}_{\vec{W}_{f/C}} = 0 \qquad (8)$$

where $\vec{C}_{b-f_{/f}}$, $\vec{M}_{\vec{R}_{/C}}$, and $\vec{M}_{\vec{W}_{f_{/C}}}$ are the ankle muscle moment, reaction force moment, and weight of the feet moment with respect to point C, respectively. By rearranging equation (8), the ankle moment can be expressed as

$$\vec{C}_{b-f_{/f}} = -\vec{M}_{\vec{R}_{/C}} - \vec{M}_{\vec{W}_{f_{/C}}} \tag{9}$$

Afterwards, the summation of moments for the whole body less the feet is taken at the point C and again for each oscillation

$$\vec{C}_{f_{/b-f}} + \vec{M}_{\vec{W}_{b-f_{/C}}} = I_{b-f_{/C}} \cdot \alpha_{b-f} \tag{10}$$

where $\vec{M}_{\vec{W}_{b-f_{/C}}}$, $I_{b-f_{/C}}$ and α_{b-f} are the moment of the body weight less the feet, the moment of inertia of the body less the feet, and the angular acceleration of the whole body less the feet, respectively. Since $\vec{C}_{b-f_{/f}} = -\vec{C}_{f_{/b-f}}$ and combining equations (9) and (10) the unknown ankle moment is eliminated

$$\vec{M}_{\vec{R}_{/C}} + \vec{M}_{\vec{W}_{f_{/C}}} + \vec{M}_{\vec{W}_{b-f_{/C}}} = I_{b-f_{/C}} \cdot \alpha_{b-f} \tag{11}$$

In equation (11), the moment of inertia of the whole body less the feet is the only unknown since the moments can be estimated from anthropometric tables, video-based system and force-plate as described below.

The second method calls for the angular momentum equation of the body less the feet, with $(H_{b-f/C})$ the norm of the angular momentum of the whole body less the feet about point C in the plane of oscillation. It is expressed by

$$H_{b-f/C} = \left\| I_{b-f/COM_{b-f}} \cdot \omega_{b-f/COM_{b-f}} + \vec{r} \times \left(m_{b-f} \cdot \vec{v}_{COM_{b-f}} \right) \right\| \quad (12)$$

where $I_{b-f/COM_{b-f}}$ and $\omega_{b-f/COM_{b-f}}$ are the whole body less the feet MOI and angular velocity at its COM, respectively. $\vec{r}$ is the distance between point C and the COM of the body less the feet which is assumed to be constant, (m_{b-f}) is the mass of the body less the feet calculated by the force-plate, and $(\vec{v}_{COM_{b-f}})$ is the body COM linear velocity vector. In equation (12), the angular momentum and MOI of the body less the feet are unknown. To solve this equation, the integration of the moments at the point C, $(M_{b-f/C})$, obtained from the force-plate is calculated in the plane of oscillation

$$H_{b-f/C} = \int M_{b-f/C} \, dt \quad (13)$$

Equation (13) is then substituted into (12) to calculate the MOI values of the body less the feet about its COM by

$$I_{b-f/COM_{b-f}} = \left\| \frac{H_{b-f/C} - \vec{r} \times \left(m_{b-f} \cdot \vec{v}_{COM_{b-f}} \right)}{\omega_{b-f/COM_{b-f}}} \right\| \quad (14)$$

3.2.2.1 Data collection to calculate MOI of the whole body less the feet by the inverse dynamics and angular momentum methods

To apply the inverse dynamics and angular momentum approaches, the mass of the whole body less the feet and its COM positions during the self-imposed oscillations, ground reaction forces, COP, position of the point *C*, moments at the point *C*, along with the feet mass and its COM locations are required. The personalized procedures with minimum used of anthropometric data to calculate these parameters are presented below.

The whole body COM position along the longitudinal axis of the body was calculated by the reaction board method (Winter, 2005). The subjects were asked to lie down in supine position on a rigid board mounted on a force-plate at one end and a pivot point at the other end. Then, the ground reaction forces were recorded by the force-plate for a period of 5 s and the mean values were computed. Knowing the weight of the balance board (w_1), its COM location from the pivot (x_1), the body weight (w_2), the distance between force-plate and the pivot (x_3), and force-plate reading (S) in Newton with the body lying supine on the board, the body COM position along its longitudinal axis (x_2) was calculated by the second equation of equilibrium ($x_2 = \frac{Sx_3 - w_1x_1}{w_2}$). Afterwards, the distance between the whole body COM less the feet with respect to the point *C* was calculated by subtracting the lateral malleoli height from the whole body COM position. The mass of the body less the feet was calculated by subtracting the feet mass from the measured whole body mass using anthropometric tables (de Leva, 1996). This is the only time that data from anthropometric tables are utilized.

The horizontal positions of the whole body COM were obtained from force-plate data when the subject stands upright in a quiet position. Subjects were instructed to stand in the middle of the force platform with arms beside the trunk and feet were parallel and 10 cm apart from each other. A quiet standing trial of 20 s at 60 Hz was recorded to calculate the mean position of the COP. This mean value of the COP corresponds to the position of whole body COM (Murray et al., 1967). Then, by subtracting the horizontal COM positions of the feet, as described below, from those of the whole body using equation (1), the COM positions of the whole body less the feet along AP and ML axes were obtained.

A video-based system and a force-plate were used to calculate the linear and angular velocities and accelerations of the whole body less the feet COM, the ground reaction forces, the COP coordinates during the oscillations, the point *C*, moments at the point *C*, and the position of the feet COM. Six cameras (Motion Analysis Corporation, Santa Rosa, USA) were located around a force-plate at a distance of about 3 m from its center. Fifteen retro reflective markers of 16 mm in diameter were applied. Four of them were put over the front, back and sides of the trunk aligned to the mid-sagittal and mid-frontal planes of the body at the height of the previously calculated COM. These markers were used for locating and tracking the 3-D coordinates of the COM of the whole body less the feet during the self-imposed oscillations. Four other markers were put over the right and the left lateral malleoli and the second metatarsophalageal joints to define the point C and feet COM. The horizontal positions of the feet COM were calculated at a midpoint between the malleoli and the heads of the second metatarsals and used to calculate the horizontal positions of the whole body less the feet COM as described above. One concern was to ensure that subjects moved as a rigid block. To verify this assumption, seven additional markers were fixed to over the top of the head, acromia, greater trochanters, and lateral epicondyles of the femur.

Subjects were instructed to stand in the middle of the force platform with arms beside the trunk and feet were parallel and 10 cm apart from each other. First, a quiet standing trial of 20 s was obtained while video and force-plate data were recorded simultaneously. Afterwards, they were asked to perform three sets of five trials of self-imposed oscillations. Subjects performed AP and ML oscillations of about 20° in amplitude then a rotation of 40° in amplitude about the LG axis of the body as illustrated in Figure 3.4. For all acquisitions, video and force-plate data were collected simultaneously at 60 Hz for a 20 s period.

Afterwards, video (3-D markers' coordinates) and force-plate data were filtered with a fourth-order zero-phase lag Butterworth filter having a cutoff frequency of 6 Hz (Allard et al., 1995; Carpenter, 2001). The 3-D coordinates of the markers were calculated in three steps. First, a preliminary calculation of the cameras' positions was performed using a seed calibration by means of direct linear transformation (Abdel- Aziz and Karara, 1971) method. Then, a wand with three markers was waved around throughout the capture volume for 60 s to generate 10800 calibration points. Finally, the 3-D coordinates of the markers were calculated from two-dimensional image coordinates by the Motion Analysis Corporation's software using the bundle adjustment, where multiple images of the same point were applied for positioning, orientation, and calibration of the cameras (Gruen, 1997).

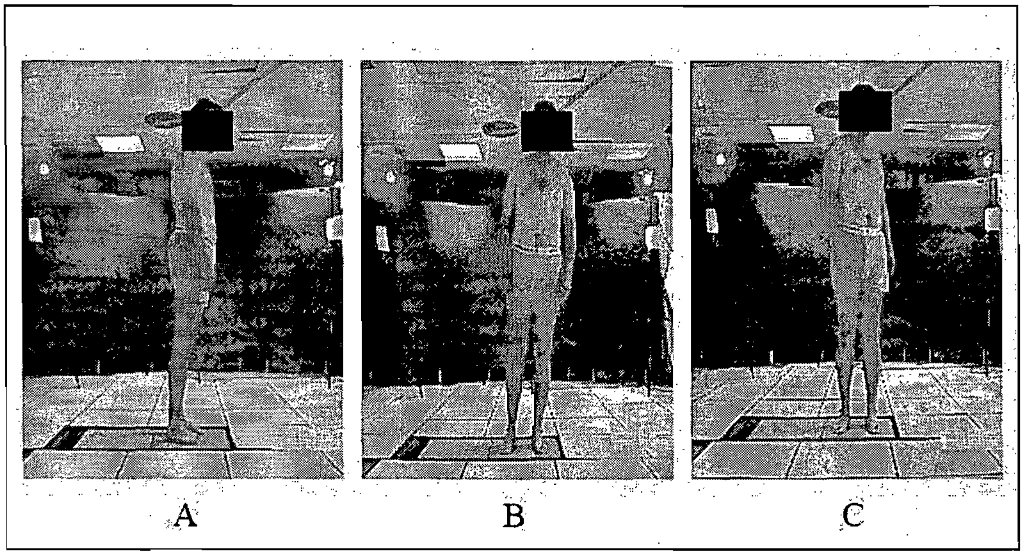

Figure 3.4. Illustration of the three self-imposed oscillations of the whole body less the feet on: A) antero-posterior, B) medio-lateral, and C) longitudinal axes through a mid-point between the ankles.

Using the head, acromia, greater trochanters, and lateral epicondyles of the femur makers, the average relative displacements of the head, trunk, thighs, and legs were less than 3 mm for all three rotations. Thus, we can assume that the whole body less the feet has oscillated as a rigid segment during the trials. These markers were also used to calculate maximal out of plane deviations during the oscillations. For the AP and ML axes this corresponded to 1.2° off-set while for the LG axis it was less than 1.8° in the sagittal and frontal planes. A trial was excluded from the subsequent analyses if the average relative displacements and the out of plane deviations of the whole body were above these values.

From the quiet standing trial, the positions of the point *C*, the whole body and feet COM in AP and ML directions were computed. The position of the point *C* was calculated as the mid-point between the lateral malleoli. The whole body and feet COM in AP and ML directions were used to compute the correspondingly coordinates of the whole body less the feet COM as described above. These results were then applied to determine the distances between the whole body COM less the feet and the four markers put on the trunk at the height of the previously calculated COM. The distances between the markers and the body less the feet COM were required to track the COM during the self-imposed oscillations.

Afterwards, the MOI for the whole body less feet about the midpoint of the ankles were estimated by the inverse dynamics method using equation (11) and then transposed to the COM of the whole body less the feet. Angular momentum equation (equation 14) was also employed to estimate MOI of the whole body less the feet about its COM.

3.3 Moment of inertia of the head-neck-trunk segment

The MOI values of the head-neck-trunk segment were estimated by anthropometric method of de Leva (1996), Hanavan's geometrical model (1964), the photogrammetric method of Jensen (1978), and angular momentum approach. Inverse dynamics method was not applied to estimate the MOI values of head-neck-trunk because of its larger variability and inaccuracies to calculate moments at the joint centers of the oscillating segments. With the angular momentum method the mass and COM locations of head-neck-trunk are needed to calculate the segment's MOI values. To avoid using anthropometric tables for these values, a new force-plate technique was developed and presented below.

3.3.1 Anthropometric, geometric, and photogrammetric methods

For the head-neck-trunk segment's MOI values the same procedure as the whole body less the feet was applied using the regression equations of de Leva (1996). The head-neck-trunk mass was calculated by summation of the sub-segment masses (i.e., head-neck and trunk). The trunk and head-neck segments' COM positions with respect to the proximal (hip) joint center were calculated in a global coordinate system. The origin of this coordinate system was located in a mid-point between the two greater trochanters. Afterwards, the 3-D COM coordinates of the head-neck-trunk segment were calculated using equation (1). Using the radii of gyration with respect to the each segment length, the MOI values along the three axes of rotation through the COM were estimated. Then, the head-neck-trunk segment MOI values about its COM for all the three principal axes were calculated by means of parallel axis theorem.

The head, neck, and trunk COM and MOI values along the three principal axes were calculated with respect to each segment's dimensions using the Hanavan (1964) and Jensen (1978) models. After estimating of SIP for these segments, the same approach as described for the de Leva (1996) method was followed to estimate the head-neck-trunk inertial properties.

3.3.2 Angular momentum method

With the angular momentum method to estimate the head-neck-trunk segment's MOI values, the segment's mass and COM location must be determined. A new force-plate technique was developed to measure segment masses and COM positions along their longitudinal axis as follow to avoid using anthropometric data from de Leva anthropometric methods or others. First, this new method will be described and then the application of the angular momentum method to estimate the head-neck-trunk MOI values.

3.3.2.1 Calculations to estimate the segments' mass and COM position using a new force-plate technique

The technique is based on changes in the force-plate moments due to a segment (e.g., right upper limb) displacements can be used to determine either its mass or COM position. While a subject stands quietly on a force-plate with arms beside the body as illustrated in Figure 3.5, the product of the mass of the whole body by its COM position $(m_b \bullet x_{com_b})$ along the AP axis with respect to force-plate center is equal to

$$m_b(x_{com_b})_i = m_{b-s}(x_{com_{b-s}})_i + m_s(x_{com_s})_i \qquad (15)$$

where m_b, m_{b-s}, and m_s are the whole body mass, whole body less the displaced segment (e.g., right upper limb) mass, and the displaced segment mass, respectively, and x_{com_b}, $x_{com_{b-s}}$, and x_{com_s} are the horizontal positions of the COM of the whole body, whole body less the displaced segment, and displaced segment along the AP axis, respectively.

Then, if the right upper limb is flexed at 90° the product of the mass of the whole body by its COM position becomes

$$m_b(x_{com_b})_f = m_{b-s}(x_{com_{b-s}})_f + m_s(x_{com_s})_f \qquad (16)$$

where subscripts i and f refer to initial and final positions (i.e. prior to and following limb displacement) of the right upper limb, respectively. Since $m_{b-s}(x_{com_{b-s}})_i = m_{b-s}(x_{com_{b-s}})_f$ and combining the equations (15) and (16), the mass of the displaced segment is calculated by

$$m_s = m_b\left[\frac{(x_{com_b})_f - (x_{com_b})_i}{(x_{com_s})_f - (x_{com_s})_i}\right] \qquad (17)$$

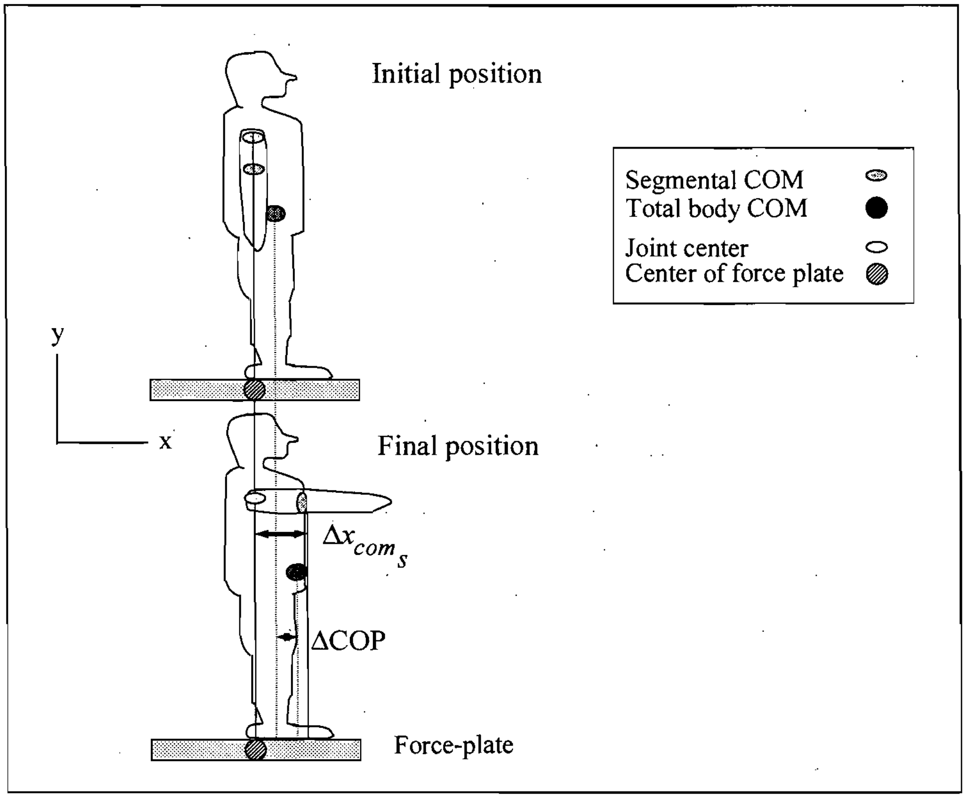

Figure 3.5. Illustration of the change in center of pressure as related to the change in segmental center of mass location during quiet standing on the force-plate.

If in the initial position the horizontal location of the displaced segment (right upper limb) COM coincides with the center of the force-plate along its AP axis, then $(x_{com_s})_i = 0$. By definition:

$$\Delta\text{COP} = (x_{com_b})_f - (x_{com_b})_i \qquad (18)$$

Thus, the COM position of the displaced segment is calculated by

$$(x_{com_s})_f = \frac{m_b(\Delta\text{COP})}{m_s} \qquad (19)$$

In equation (19) the displaced segment mass (m_s) is unknown, thus the COM position of the displaced segment (x_{com_s}) cannot be calculated. To solve the problem, the subject is required to assume the same body segment configuration while lying on a reaction board as described below.

For a subject lying on a reaction board as shown in Figure 3.6 two readings of the force-plate are made, namely, with the segment to be measured held in two different positions. In the initial position the desired segment (e.g., right upper limb) is extended and kept in neutral position, while in final position it is lifted to a vertical position so that its COM lies over the related joint center. The segment mass can be calculated by

$$m_s = \frac{(\mathrm{S}'-\mathrm{S})x_b}{x_j - x_{com_s}} \qquad (20)$$

where S' and S are force-plate readings following, and prior to the limb displacement, respectively, and x_b, and x_j are the distance between the knife edges (i.e., pivot point and the installation point of the board to the force-plate as shown in Figure 3.6.) and the distance between joint center of rotation and the pivot point, respectively. By substituting equation (19) on (20) the displaced segment mass is

$$m_s = \frac{m_b \Delta\mathrm{COP} + (\mathrm{S}'-\mathrm{S})x_b}{x_j} \qquad (21)$$

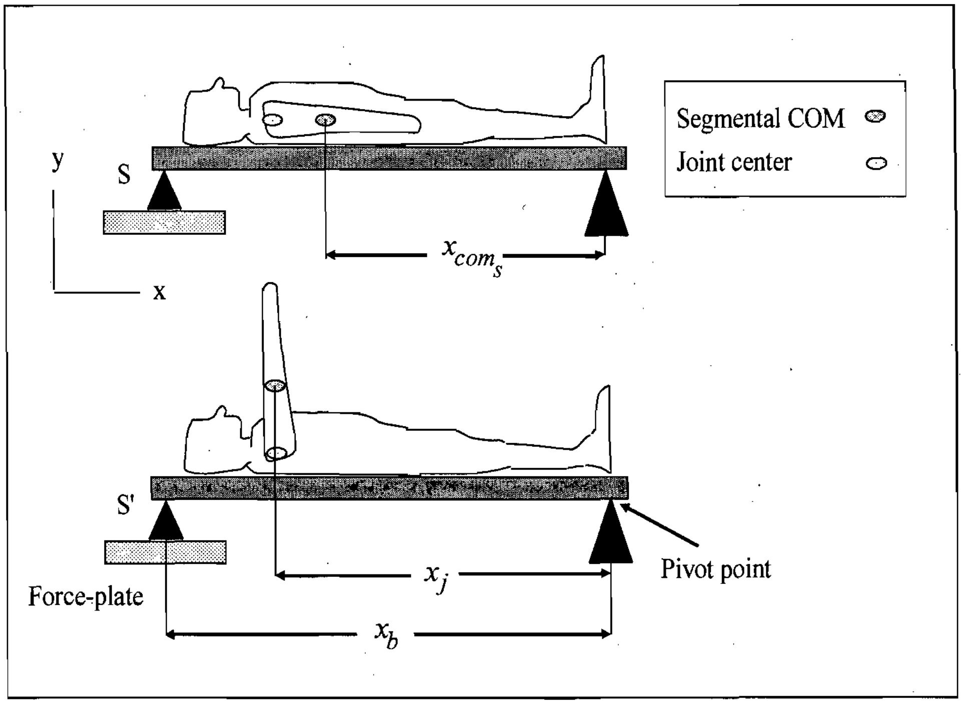

Figure 3.6. Illustration of the change in moments as related to the change in segmental center of mass location during lying on the reaction board.

By substituting m_s in equation (20) the COM position of the shifted segment is calculated from the pivot point by

$$x_{com_s} = x_j - \frac{(S'-S)x_b}{m_s} \qquad (22)$$

3.3.2.2 Data collection for the new force-plate technique to calculate segments' mass and COM position

The subjects were asked to move only the displaced segment while keeping the other body segments in neutral position. Small movements during the total leg flexion occurred because it: (1) limited pelvic movement, (2) prevented lumbar flatting, and (3) was within a comfortable range of motion. Therefore, these small movements are assumed to have negligible effect on the total leg's mass and COM position.

The upper and lower limbs and their sub-segments (i.e., forearm and hand, and leg and foot) were placed similarly in both quiet standing on the force-plate and then in the lying position on the board. The body segments displacements were monitored visually. Only the right limbs were measured. For quiet standing six segment configurations were tested:

1. Subjects were asked to stand quietly while ankle, knee, and hip of the right limb coincided with the center of the force-plate. The left foot was parallel to and at a distance of 10 cm from the right foot and both upper limbs were positioned along the trunk,
2. 90° knee flexion of the right limb,
3. Right hip flexion to 90° with the thigh paralleled to the force-plate surface,
4. Subjects were asked to stand quietly while wrist, elbow, and shoulder of the right limb coincided with the center of the force-plate. Again the left foot was parallel to and at a distance of 10 cm from the right foot and both upper limbs were beside the trunk,
5. Elbow flexion of the right limb to 90°, and
6. Right shoulder flexion to 90° was achieved such that the palm of the right forearm was parallel to the force-plate surface.

After that, subjects were asked to lie down on a reaction board. For this position, the following six segment configurations were tested:

1. Subjects were asked to lie in prone position on the board. The legs and arms were extended (neutral position) and the upper limbs were beside the trunk,
2. Knee flexion of the right limb to 90°,
3. Afterwards subjects were asked to lie in supine position on the board. The legs and arms were extended (neutral position) and the upper limbs were beside the trunk,
4. Right hip flexion to 90° was made such that the right sole was parallel to the reaction board surface,
5. Elbow flexion of the right limb to 90°, and
6. Upper limb flexion (shoulder flexion) to 90° with the right limb perpendicular to the reaction board surface was achieved.

The force-plate data were collected at 60 Hz for a period of 5 s for all the trials. The mean values of COP positions and reaction forces were calculated. Afterwards, the masses and COM positions of the upper and lower limbs, forearm and hand, and leg and foot segments were calculated using equations (21) and (22). Then, the upper arm and thigh segments' masses were calculated as the difference between those of the upper limb and forearm-hand, and lower limb and leg-foot, respectively. The mass of the head-neck-trunk segment was calculated as the difference between the whole body mass (m_b) and twice of the upper and lower limbs' mass. After that, the upper arm and thigh COM positions with respect to the relative proximal endpoints were calculated using equation (1). Knowing the whole body, upper and lower limbs masses and their COM positions with respect to the pivot point (Figure 3.6.), the COM position of the head-neck-trunk segment with respect to that point and the mid-hip joint centers (along the longitudinal axis) can be calculated by equation (1). To be comparable with the other methods, the segment's endpoints were chosen similar to those of the previous studies.

3.3.2.3 Data collection for the angular momentum method to calculate head-neck-trunk segment's MOI values

To apply the angular momentum method to calculate the head-neck-trunk segment's MOI values from those of the whole body less the feet, the MOI values of the upper and lower limbs are needed. Thus, the upper and lower limbs masses and their 3-D COM positions are required. The procedure to obtain these values follows.

To obtain the horizontal positions of the upper and lower limbs COM with respect to the whole body less the feet COM locations, four reflective markers were used. These markers were put over the right upper and lower limbs calculated COM positions and aligned with respect to their mid-sagittal and mid-frontal planes as shown in Figure 3.7. Since symmetry of the right and left segments were assumed, only the right limbs were considered. The intersections of these markers served as the horizontal COM positions of the upper and lower limbs. Five markers were also put over the lateral malleoli, right humeral head, and greater trochanters to define the point C (midway between the lateral malleoli), right shoulder, and right hip center, respectively. The position of point C is required to determine the left upper and lower limbs' COM positions in the horizontal directions since it is assumed to be located in the mid-way between the right and left limbs' COM positions. The right hip center was assumed to be located at 25% of the distance between the right and left greater trochanters markers from the right side marker (Robertson et al., 2004).

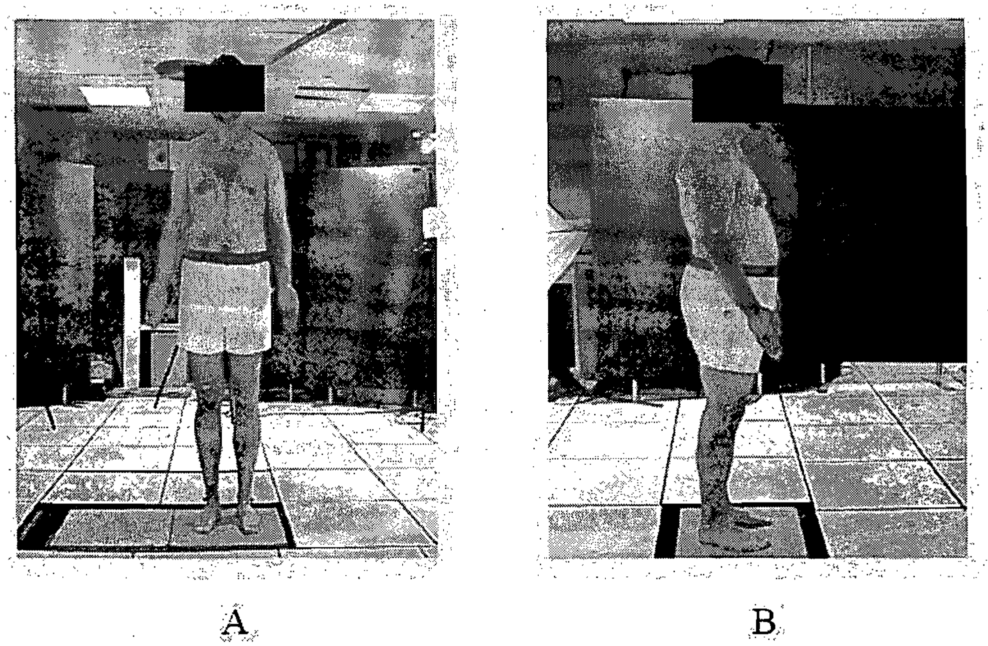

Figure 3.7. Illustration of the markers put over the right upper and lower limbs calculated COM positions, malleoli, right humeral head, and greater trochanters to determine the horizontal positions of the upper and lower limbs.

Once the reflective markers were in place, subjects were instructed to stand in the middle of the force-plate with upper limbs beside the trunk and feet parallel and 10 cm apart from each other. Afterwards, a quiet standing trial of 20 s using the video-based system and a force-plate was recorded. Knowing the horizontal COM positions of the body, and the upper and lower limbs, the head-neck-trunk segment's COM positions along AP and ML axes can be calculated by equation (1). The markers put over the right upper and lower limbs were also used to calculate the linear and angular velocities of these segments during their self-impose oscillations.

Three sets of the self-imposed oscillations of the upper and lower limbs were performed about AP, ML, and LG axes of the shoulder and hip, respectively. Subjects performed AP and ML oscillations of about 90° in amplitude then a rotation of 45° in amplitude about the LG axis of the relative proximal joint centers as illustrated in Figure 3.8. Five trials were performed for each axis of rotation and the mean MOI values were calculated. For all acquisitions, video and force-plate data were collected simultaneously at 60 Hz for a 20 s period. Afterwards, the data were filtered with a fourth-order zero-phase lag Butterworth filter having a cutoff frequency of 6 Hz. Particular care was taken to make sure that the upper and lower limbs oscillate as a rigid body and the out of plane deviations is minimal during the oscillations. For the AP and ML axes the out of plane deviations of the upper and lower limbs corresponded to 1.4° and 1.6° off-set, respectively. For the LG axis the maximum deviations were 1.3° and 1.9° for the sagittal and frontal planes, respectively. A trial was excluded if the average out of plane deviations of the upper and lower limbs were above these values.

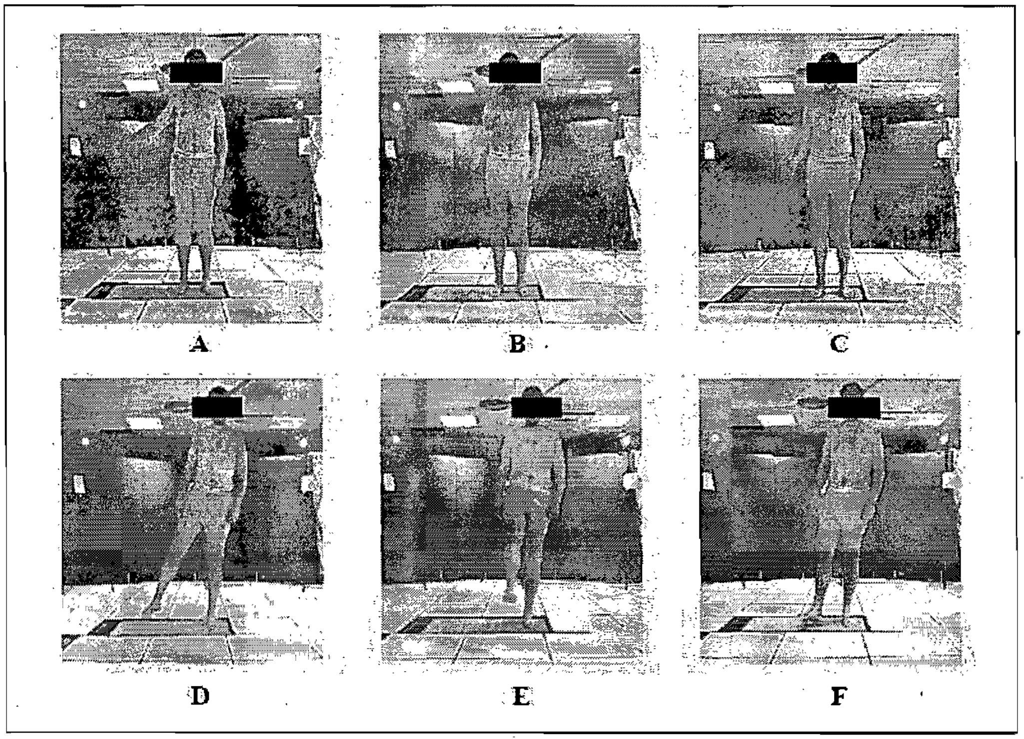

Figure 3.8. Illustration of the three self-imposed oscillations of the upper and lower extremities about: A and D) antero-posterior, B and E) medio-lateral, and C and F) longitudinal axes through the related proximal joint centers.

The upper and lower limbs' MOI values with respect to their COM positions were calculated using the angular momentum equation

$$I_{l_{/COM}} = \left\| \frac{H_{l_{/ps}} - \vec{r}_l \times \left(m_l \cdot \vec{v}_{l_{/COM}} \right)}{\omega_{l_{/COM}}} \right\| \quad (23)$$

where $I_{l_{/COM}}$, $H_{l_{/ps}}$, $\vec{r}_l$, m_l, $\vec{v}_{l_{/COM}}$, and $\omega_{l_{/COM}}$ are the limbs' MOI values with respect to their COM, angular momentum of the oscillating limbs at their proximal joint, the distance between the limbs' COM location and the relative proximal joint centers along the three principal axes, limb masses, and linear and angular velocities of the limbs' COM during the self-impose oscillations, respectively. Since the whole body was fixed except for the oscillating limb, the difference between the integration of moments (i.e., angular momentum) of the whole body less the feet during quiet standing and the limb oscillation was considered as the angular momentum of the oscillating limb. The angular momentum of the oscillating limb was corrected by eliminating the off-set resulting from the angular momentum measured during quiet standing. This off-set was about 0.008 kg*m^2/s comparatively to 0.572 kg*m^2/s during an oscillation. The mean of the five trials performed along each axis of rotation was used to calculate the MOI values.

The feet MOI values was estimated by de Leva (1996) method and subtracted from those of the lower limb using parallel axis theorem. After that, the head-neck-trunk segment's MOI values along the three principal axes of rotation about its COM were computed as the difference between the whole body MOI less the feet estimated by the angular momentum method and twice the sum of the limbs' MOI given by equation (2).

3.4 Data analysis

Within each of the three morphologic groups repeated measures ANOVA were used to compare the MOI values of the whole body less the feet and the head-neck-trunk segment estimated by the methods and for each principal axis.

This was followed by a post-hoc analysis using the Bonferroni test if a statistical difference was observed ($\alpha = 0.05$). Since the mass and COM location of the head-neck-trunk segment estimated by different methods have an important effect on the segment's MOI values, these were compared within each morphological group using repeated measures ANOVA. Again, these were followed by a post-hoc analysis using Bonferroni adjustments where a statistical difference was observed ($\alpha = 0.05$).

To verify if the results of the MOI values of the whole body less the feet and the head-neck-trunk inertial parameters provided by a method behaved similarly within each morphologic group, these values were arbitrary compared to those obtained with the de Leva (1996) method (as a criterion) by means of Pearson coefficients of correlation. The percent differences between the de Leva (1996) method and the other methods were calculated.

Chapter 4

4. RESULTS

This chapter begins with the results of the whole body less the feet moment of inertia values. This is followed by the head-neck-trunk segment mass and its COM locations. These were included here because to calculate head-neck-trunk MOI values from the whole body MOI less the feet, the segment mass and COM position were needed. Then, the head-neck-trunk moments of inertia along all the three principal axes at its COM and for each morphological group are presented. Since lower and upper limbs masses and their COM positions are intermediate results, they are presented in appendix B.

4.1 Whole body less the feet moment of inertia results

Whole body less the feet MOI values for the three morphological groups and all the three principal axes are presented in four parts. The first part reports on the MOI values for the normal BMI range. This is followed by the MOI values for the lean and obese BMI groups. Then, the Pearson coefficients of correlation between the de Leva (1996) method (as a reference) and the other approaches for all three morphological groups and axes are presented.

4.1.1 Whole body less the feet MOI of the normal BMI range

Results for the whole body less the feet MOI for the normal BMI group are presented in Table 4.1. Repeated measures ANOVA showed no significant difference ($p<0.05$) for the normal BMI range among all methods for all three principal axes. For the AP axis the average moment of inertia was 11.99±0.45 kgm^2 while for the ML axis it was 11.39±0.53 kgm^2 and 1.30±0.06 kgm^2 for the LG axis. For this group of subjects the average values of the inverse dynamics and angular momentum methods were within the MOI ranges, while those of Hanavan (1964) and Jensen (1978) models were the lowest and highest values, respectively. The average MOI values estimated by Hanavan method (1964) were the closest (−1.2%) to those of de Leva (1996) while the photogrammetric method yielded at the highest average difference (+9.8%). The MOI values of inverse dynamics and angular momentum methods were in average 7.3 and 6.4% higher than those of the de Leva (1996), respectively.

Table 4.1. Means, standard deviations, and ranges of the moment of inertia ($kg.m^2$) for the whole body less the feet calculated at the center of mass of subjects with a normal BMI along with the % difference between de Leva (1996) method and the other methods. Data are presented for antero-posterior (AP), medio-lateral (ML), and longitudinal (LG) axes

Method	Axis	Mean	S.D.	Range	% difference
de Leva (1996)	AP	11.46	1.67	9.03-13.22	–
	ML	10.97	1.44	8.73-12.90	–
	LG	1.26	0.14	1.03-1.41	–
Hanavan (1964)	AP	11.65	1.84	9.13-13.89	1.7
	ML	10.66	1.43	8.64-12.74	−2.8
	LG	1.23	0.17	1.06-1.52	−2.5
Jensen (1978)	AP	12.58	1.65	10.60-14.88	9.8
	ML	11.88	1.39	9.95-13.85	8.3
	LG	1.28	0.20	1.04-1.49	1.6
Inverse dynamics	AP	12.21	1.29	10.72-14.08	6.5
	ML	11.76	1.40	10.33-13.97	7.0
	LG	1.37	0.21	1.15-1.79	8.5
Angular momentum	AP	12.10	1.26	10.58-13.77	5.7
	ML	11.67	1.19	10.38-13.22	6.4
	LG	1.35	0.26	0.99-1.80	7.1

%Difference = (other mean – de Leva mean)*100/de Leva mean

4.1.2 Whole body less the feet MOI of the lean group

Values for subjects in the lean BMI group are presented in Table 4.2. The de Leva (1996) method gave generally the highest MOI values. The lowest values were found with the inverse dynamics and angular momentum methods while the other two methods provided values close to them. Repeated measures ANOVA showed statistically significant differences in the MOI values ($p<0.05$). The MOI values based on the de Leva method (1996) were higher by 13.3% ($p=0.019$) and 12.9% ($p=0.012$) for the AP axis, and 16.4% ($p=0.02$) and 15.8% ($p=0.019$) for the ML axis from those of the inverse dynamics and angular momentum methods, respectively. The values based on Hanavan's model (1964) were also higher by 14.1% ($p=0.034$) and 13.5% ($p=0.035$) along the ML axis compared to the inverse dynamics and angular momentum methods, respectively. For the LG axis, the MOI data based on the de Leva (1996) and Hanavan (1964) methods were 9.6% ($p=0.018$) and 14.7% ($p=0.021$) higher than those of the inverse dynamics approach.

Table 4.2. Means, standard deviations, and ranges of the moment of inertia ($kg.m^2$) for the whole body less the feet calculated at the center of mass of subjects with a lean BMI along with the % difference between de Leva (1996) method and the other methods. Data are presented for antero-posterior (AP), medio-lateral (ML), and longitudinal (LG) axes

Method	Axis	Mean	S.D.	Range	% difference
de Leva (1996)	AP	10.29	2.11	6.39-12.82	–
	ML	10.08	2.17	5.72-12.22	–
	LG	0.94	0.12	0.69-1.07	–
Hanavan (1964)	AP	10.07	2.12	5.69-12.17	−2.1
	ML	9.81	2.08	5.58-11.99	−2.6
	LG	0.98	0.13	0.72-1.12	4.6
Jensen (1978)	AP	9.07	1.87	6.17-11.21	−11.8
	ML	8.90	2.11	6.11-12.38	−11.7
	LG	0.80	0.24	0.48-1.03	−14.0
Inverse dynamics	AP	8.92*	2.31	5.13-11.25	−13.3
	ML	8.43*§	2.12	4.94-10.71	−16.4
	LG	0.85*§	0.16	0.50-0.99	−9.6
Angular momentum	AP	8.96*	2.33	4.98-11.25	−12.9
	ML	8.49*§	2.09	4.99-10.59	−15.8
	LG	0.83	0.14	0.51-1.00	−11.7

Significant differences between methods in each group are shown by superscripts ($p<0.05$),
*de Leva vs. the other methods,
§ Hanavan vs. the other methods,
%Difference = (other mean – de Leva mean)*100/de Leva mean

4.1.3 Whole body less the feet MOI of the obese group

Table 4.3 represents values for subjects in the obese BMI group. The de Leva (1996) method gave generally the lowest MOI values. The inverse dynamics and angular momentum methods had the highest values. Generally, statistical differences were noted between the de Leva (1996) and the Hanavan (1964) methods and those of inverse dynamics and angular momentum.

The MOI values based on the de Leva method (1996) were lower by 25.0% (p=0.011) and 24.3% (p=0.01) for the AP axis, and 18.6% (p=0.014) and 17.9% (p=0.005) for the ML axis from those of the inverse dynamics and angular momentum methods, respectively. These values based on Hanavan's model (1964) were lower by 18.3% (p=0.013) and 17.6% (p=0.003) for the ML axis compared to the inverse dynamics and angular momentum methods, respectively. For the LG axis, the MOI values based on the de Leva (1996) and Hanavan (1964) methods were 10.2% (p=0.023) and 16.1% (p=0.01) lower than those of the inverse dynamics method. Significant difference was also observed along the LG axis between the de Leva (1996) method and Hanavan's model (1964) for the MOI values (p=0.017).

Table 4.3. Means, standard deviations, and ranges of the moment of inertia (kg.m^2) for the whole body less the feet calculated at the center of mass of subjects with an obese BMI along with the % difference between de Leva (1996) method and the other methods. Data are presented for antero-posterior (AP), medio-lateral (ML), and longitudinal (LG) axes

Method	Axis	Mean	S.D.	Range	% difference
de Leva (1996)	AP	15.32	2.22	12.20-18.44	–
	ML	14.69	2.05	11.99-18.02	–
	LG	1.77	0.12	1.56-1.96	–
Hanavan (1964)	AP	15.78	2.17	12.03-18.11	3.0
	ML	14.73	1.84	11.27-16.77	0.2
	LG	1.68*	0.10	1.51-1.84	–4.9
Jensen (1978)	AP	15.48	0.97	13.64-16.66	1.0
	ML	14.22	0.87	12.29-14.98	–3.2
	LG	1.77	0.31	1.44-2.28	–0.1
Inverse dynamics	AP	19.15*	3.91	14.65-25.88	25.0
	ML	17.43*§	2.71	13.07-20.81	18.6
	LG	1.95*§	0.17	1.80-2.23	10.2
Angular momentum	AP	19.05*	3.82	14.75-25.39	24.3
	ML	17.32*§	2.30	13.74-20.39	17.9
	LG	1.91	0.24	1.42-2.14	7.9

Significant differences between methods in each group are shown by superscripts ($p<0.05$),
*de Leva vs. the other methods,
§ Hanavan vs. the other methods,
%Difference = (other mean – de Leva mean)*100/de Leva mean

4.1.4 Pearson coefficients of correlation of the whole body less the feet MOI

The Pearson coefficients of correlation between the de Leva (1996) and the other methods for all three morphological groups and axes are given in Table 4.4. The overall average coefficient of correlation was 0.76±0.31. As a group, subjects with a lean BMI displayed the best average correlation (0.91). The worst absolute average correlations were found for the photogrammetric method of Jensen (0.64) and along the longitudinal axis (0.42). This method was the only one with negative correlations. The MOI values obtained from the Hanavan (1964), inverse dynamics and angular momentum methods were well correlated to those of de Leva (1996) for all the three morphological groups (0.82, 0.85, and 0.83, respectively). The proposed methods were also more strongly correlated to the de Leva method values for the lean subjects compared to the two other morphological groups.

Table 4.4. Pearson coefficients of correlation (*p* values) of the moment of inertia estimations between de Leva method and the other approaches for the normal, lean, and obese morphological groups about the antero-posterior (AP), medio-lateral (ML), and longitudinal (LG) axes

Method	Axis	Normal	Lean	Obese
Hanavan (1964)	AP	0.97 (0.000)	0.90 (0.003)	0.84 (0.010)
	ML	0.97 (0.000)	0.97 (0.000)	0.67 (0.067)
	LG	0.17 (0.721)	0.98 (0.000)	0.91 (0.001)
Jensen (1978)	AP	0.90 (0.006)	0.89 (0.003)	0.59 (0.128)
	ML	0.83 (0.020)	0.85 (0.007)	0.48 (0.255)
	LG	−0.38 (0.405)	0.81 (0.015)	−0.07 (0.870)
Inverse dynamics	AP	0.92 (0.003)	0.94 (0.001)	0.92 (0.001)
	ML	0.91 (0.004)	0.90 (0.003)	0.83 (0.010)
	LG	0.44 (0.324)	0.98 (0.000)	0.79 (0.020)
Angular momentum	AP	0.94 (0.001)	0.95 (0.000)	0.93 (0.001)
	ML	0.83 (0.022)	0.91 (0.002)	0.85 (0.007)
	LG	0.33 (0.467)	0.81 (0.016)	0.93 (0.001)

4.2 Head-neck-trunk segment's mass and COM position

Head-neck-trunk mass and COM position for the three morphological groups are presented in three parts. The first part deals with the mass data for the head-neck-trunk segment followed by the COM position. Then, the Pearson coefficients of correlations between the de Leva (1996) method and the other approaches for all three morphological groups are presented.

4.2.1 Head-neck-trunk segment mass

Table 4.5 presents the mean values of head-neck-trunk mass for all the three morphological groups. The de Leva (1996) method provided the lowest masses, while the Dempster (1996) method generally gave the highest values. Repeated measures ANOVA showed statistically significant differences on the segment mass estimations. Post-hoc analyses on the mass data showed significant differences between the de Leva (1996) method and all the other methods for each morphological group except for the Jensen's method (1978) for the normal BMI group ($p<0.05$). The de Leva (1996) values were lower than those of the other methods by 6.3 to 16.2%. The head-neck-trunk mass values based on Dempster (1955) were statistically higher ($p<0.05$) than those of Hanavan's model (1964) for all the three groups, Jensen (1978) for lean subjects, and the force-plate technique for lean and obese groups by maximum 16.0%. Significant difference was also observed between Hanavan's model (1964) and photogrammetric method of Jensen (1978) for the lean ($p=0.04$) participants. Generally the new force plate technique and the Jensen (1978) method provided values closer to those of de Leva for all three body types compared to Dempster (1955) and Hanavan (1964) methods.

Table 4.5. Means, standard deviations (S.D.), and the ratios (S.D.) of head-neck-trunk mass (kg) as a percent of whole body mass (WBM) along with the % difference between the de Leva (1996) method and the other method. Data are presented for lean (L), normal (N), and obese (O) morphological groups

Method	Group	Mean	S.D.	%WBM	% difference
de Leva (1996)	L	26.61	4.10	49.8 (0.6)	–
	N	37.99	1.43	50.4 (0.0)	–
	O	50.40	2.14	50.4 (0.0)	–
Dempster (1955)	L	30.87*	4.46	57.8 (0.0)	16.0
	N	43.56*	1.64	57.8 (0.0)	14.7
	O	57.80*	2.46	57.8 (0.0)	14.7
Hanavan (1964)	L	29.81*‡	4.32	55.8 (0.0)	12.0
	N	42.11*‡	1.59	55.9 (0.0)	10.8
	O	55.90*‡	2.38	55.9 (0.0)	10.9
Jensen (1978)	L	28.29*‡§	3.77	52.9 (1.6)	6.3
	N	41.06	2.80	54.5 (5.3)	8.1
	O	58.55*	3.63	58.6 (2.0)	16.2
Force-plate technique	L	29.56*‡	4.29	55.4 (1.6)	11.1
	N	41.80*	2.54	55.5 (2.2)	10.0
	O	55.44*‡	2.94	55.4 (1.2)	10.0

Significant differences between methods in each group are shown by superscripts ($p<0.05$),
*de Leva vs. the other methods,
‡Dempster vs. the other methods,
§ Hanavan vs. the other methods,
%Difference = (other mean – de Leva mean)*100/de Leva mean

4.2.2 Head-neck-trunk segment COM position

The mean values of the head-neck-trunk COM location with respect to the hip center based on the applied methods and for each morphological group are presented in Table 4.6. Repeated measures ANOVA showed the COM locations based on the de Leva (1996) method were statistically further with respect to the segment's proximal endpoint by a maximum of 25.5% from those of Hanavan (1964) and Jensen (1978) for all the groups ($p<0.05$). The de Leva values were also significantly further than those of Dempster (1955) for lean and obese subjects, and force-plate technique for lean group maximally by 16.5% ($p<0.05$). Estimations by Dempster (1955) method were statistically further than the values of Hanavan (1964) and Jensen (1978) methods for all the three morphological groups by 23.1% ($p<0.05$). Force-plate technique gave head-neck-trunk COM locations statistically higher by 21.0% from Hanavan's model (1964) for normal and obese groups ($p<0.05$), and Jensen (1978) method for the obese subjects by 30.9% ($p=0.006$). The mean value for the obese group based on Hanavan's model (1964) was observed to be significantly different from Jensen (1978) results ($p=0.003$). The segment's COM locations obtained from the new force-plate technique were in the range of the other methods for all the morphological groups.

Table 4.6. Means and standard deviations (S.D.) of trunk-head-neck center of mass location to the hip center (cm), along with the ratios (S.D.) as a % of segment length (SL) and the % difference between the de Leva (1996) method and the other methods. Data are presented for lean (L), normal (N), and obese (O) morphological groups

Method	Group	Mean	S.D.	%SL	% difference
de Leva (1996)	L	34.97	2.10	70.6 (3.0)	–
	N	36.99	1.83	68.0 (1.6)	–
	O	37.40	0.98	68.1 (0.9)	–
Dempster (1955)	L	32.71*	3.10	66.0 (0.0)	−6.5
	N	35.88	1.36	66.0 (0.0)	−3.0
	O	36.22*	1.08	66.0 (0.0)	−3.2
Hanavan (1964)	L	27.04*‡	2.61	54.6 (1.5)	−22.7
	N	29.90*‡	0.97	55.0 (0.7)	−19.2
	O	30.17*‡	1.08	54.9 (1.4)	−19.3
Jensen (1978)	L	27.92*‡	2.21	56.3 (4.9)	−20.2
	N	29.30*‡	1.96	53.9 (3.8)	−20.8
	O	27.85*‡§	0.64	50.8 (1.4)	−25.5
Force-plate technique	L	29.21*	4.64	58.9 (5.9)	−16.5
	N	35.00§	3.30	64.4 (4.2)	−5.4
	O	36.45§†	4.06	66.4 (6.7)	−2.5

Significant differences between methods in each group are shown by superscripts ($p<0.05$),
*de Leva vs. the other methods,
‡Dempster vs. the other methods,
§ Hanavan vs. the other methods,
†Jensen vs. the other methods,
%Difference = (other mean – de Leva mean)*100/de Leva mean

4.2.3 Pearson coefficients of correlation of the head-neck-trunk mass and COM position

Table 4.7 represents the Pearson coefficients of correlation of the head-neck-trunk mass and COM estimations between the de Leva (1996) and the other methods for all three morphological groups. The absolute average coefficients of correlation for the mass and COM position were 0.93±0.12 and 0.65±0.30, respectively. As a group, the lean subjects showed the best average correlation of 0.98 and 0.73 for mass and COM values, respectively. In average, Dempster (1955) and Hanavan (1964) values were well correlated to those of the de Leva (1996) for mass (0.99) and COM position (0.87). The corresponding correlation values for the new force-plate technique were 0.87 and 0.57. The worst absolute average correlations of the head-neck-trunk mass (0.82) and COM (0.26) were found for the Jensen (1978) method, the mass of the normal BMI group (0.85), and the COM position of the obese subjects (0.56). Jensen's method (1978) was the only one with negative correlations.

Table 4.7. Pearson coefficients of correlation (*p* values) of the head-neck-trunk segment's mass and center of mass location between de Leva method and the other approaches. Data are presented for the lean (L), normal (N), and obese (O) morphological groups

Method	Groups	Mass	Center of Mass
Dempster (1955)	L	0.99 (0.000)	0.96 (0.000)
	N	1.00 (0.000)	0.88 (0.008)
	O	1.00 (0.000)	0.91 (0.002)
Hanavan (1964)	L	0.99 (0.000)	0.91 (0.002)
	N	1.00 (0.000)	0.89 (0.007)
	O	1.00 (0.000)	0.68 (0.063)
Jensen (1978)	L	0.98 (0.000)	0.39 (0.340)
	N	–0.63 (0.126)	0.05 (0.908)
	O	0.86 (0.006)	0.35 (0.392)
Force-plate technique	L	0.97 (0.000)	0.64 (0.089)
	N	0.78 (0.039)	0.76 (0.047)
	O	0.91 (0.002)	0.32 (0.439)

4.3 Head-neck-trunk moment of inertia

Head-neck-trunk MOI values for the three morphological groups and all the three principal axes are presented in four sections. First, the MOI results for the normal BMI range are reported. Then, the MOI values for the lean and obese BMI groups are presented. Afterwards, the Pearson coefficients of correlations between the de Leva (1996) method and the other methods for all three morphological groups and axes are presented.

4.3.1 Head-neck-trunk moment of inertia of the normal BMI range

As presented in Table 4.8, repeated measures ANOVA showed significant differences in the MOI results for the normal BMI range. Post-hoc analyses indicated those of Hanavan's model (1964) were significantly higher (9.4 to 21.7%) in comparison with de Leva (1996) method along all the three axes of rotation ($p<0.05$). The MOI values obtained from angular momentum method were within those of the other approaches, while de Leva (1996) and Jensen (1978) methods gave the lowest and highest values, respectively. For the AP axis the average moments of inertia was 2.45 ± 0.19 kgm^2 while for the ML axis it was 2.29 ± 0.23 kgm^2 and 0.45 ± 0.05 kgm^2 for the LG axis. The average MOI values estimated by the angular momentum method were the closest (+5.8%) to those of de Leva (1996) while the Jensen (1978) method yielded at the highest average difference (+25.4%).

Table 4.8. Means and standard deviations of head-neck-trunk moments of inertia (kg/m^2) of subjects with a normal BMI, along with radii of gyration (S.D.) as a % of the segment length (SL) and the % difference between the de Leva (1996) method and the other methods. Data are presented for antero-posterior (AP), medio-lateral (ML), and longitudinal (LG) axes

Method	Axis	Mean	S.D.	%SL	% difference
de Leva (1996)	AP	2.24	0.13	44.7 (1.1)	–
	ML	2.06	0.12	42.9 (1.1)	–
	LG	0.42	0.02	19.3 (0.3)	–
Hanavan (1964)	AP	2.45*	0.11	44.4 (0.6)	9.4
	ML	2.31*	0.11	43.2 (0.5)	12.3
	LG	0.51*	0.02	20.2 (0.7)	21.7
Jensen (1978)	AP	2.70	0.44	47.2 (5.2)	20.4
	ML	2.59	0.42	46.2 (5.2)	25.4
	LG	0.41	0.07	18.4 (1.6)	−0.4
Angular momentum	AP	2.39	0.20	44.1 (3.0)	6.6
	ML	2.19	0.22	42.2 (3.1)	6.1
	LG	0.44	0.04	18.8 (1.3)	4.8

Significant differences between methods in each group are shown by superscripts ($p<0.05$),
*de Leva vs. the other methods,
%Difference = (other mean – de Leva mean)*100/de Leva mean

4.3.2 Head-neck-trunk moment of inertia of the lean group

Table 4.9 represents the mean values of the head-neck-trunk segment's MOI of the lean subjects estimated by the methods and for all the three axes of rotation. The de Leva (1996) and Jensen (1978) methods gave the lowest and highest MOI values as for the normal group, respectively. Statistical differences were noted between the de Leva (1996) method and Hanavan (1964) along all the three axes ($p<0.05$). The MOI results provided by the de Leva (1996) method were also significantly lower than those of Jensen (1978) method along AP and ML axes by maximum 52.1%, and angular momentum approach about AP by 24.4% ($p<0.05$). Significant difference was also detected between Hanavan's model (1964) and Jensen's method (1978) results about LG axis ($p=0.01$).

Table 4.9. Means and standard deviations of head-neck-trunk moments of inertia (kg/m^2) of subjects with a lean BMI, along with radii of gyration (S.D.) as a % of the segment length (SL) and the % difference between the de Leva (1996) method and the other methods. Data are presented for antero-posterior (AP), medio-lateral (ML), and longitudinal (LG) axes

Method	Axis	Mean	S.D.	%SL	% difference
de Leva (1996)	AP	1.34	0.33	45.1 (3.1)	–
	ML	1.25	0.31	43.5 (2.7)	–
	LG	0.24	0.06	19.2 (1.1)	–
Hanavan (1964)	AP	1.53*	0.29	45.9 (4.1)	14.1
	ML	1.42*	0.29	44.2 (3.8)	13.9
	LG	0.31*	0.05	20.9 (2.3)	29.4
Jensen (1978)	AP	1.94*	0.50	51.7 (6.5)	44.8
	ML	1.90*	0.51	51.0 (6.6)	52.1
	LG	0.26§	0.05	18.8 (2.2)	5.6
Angular momentum	AP	1.67*	0.32	48.2 (5.5)	24.4
	ML	1.50	0.24	45.9 (6.7)	19.9
	LG	0.25	0.05	18.7 (2.8)	3.6

Significant differences between methods in each group are shown by superscripts ($p<0.05$),
*de Leva vs. the other methods,
§ Hanavan vs. the other methods,
%Difference = (other mean – de Leva mean)*100/de Leva mean

4.3.3 Head-neck-trunk moment of inertia of the obese group

The mean values of the head-neck-trunk segment's MOI of the obese group estimated by the methods and for all the three axes of rotation are presented in Table 4.10. Again de Leva (1996) and photogrammetric (Jensen, 1978) methods provided the lowest and highest MOI values, respectively. The MOI data estimated by the de Leva (1996) method were significantly lower than those of Hanavan (1964) and Jensen (1978) methods along all the three axes at the highest difference average of 41.5% ($p<0.05$), and angular momentum method about LG axis by 27.5% ($p=0.23$). Significant differences were also observed between Hanavan (1964) and Jensen (1978) methods, and the Jensen (1978) and angular momentum approaches along AP and ML axes ($p<0.05$).

Table 4.10. Means and standard deviations of head-neck-trunk moments of inertia (kg/m^2) of subjects with an obese BMI, along with radii of gyration (S.D.) as a % of the segment length (SL) and the % difference between the de Leva (1996) method and the other methods. Data are presented for antero-posterior (AP), medio-lateral (ML), and longitudinal (LG) axes

Method	Axis	Mean	S.D.	%SL	% difference
de Leva (1996)	AP	3.01	0.18	44.5 (0.6)	–
	ML	2.74	0.18	42.5 (1.0)	–
	LG	0.54	0.06	18.8 (0.7)	–
Hanavan (1964)	AP	3.20*	0.23	43.6 (1.0)	6.3
	ML	3.02*	0.20	42.4 (1.0)	10.2
	LG	0.74*	0.13	20.9 (1.7)	38.4
Jensen (1978)	AP	3.56*§	0.29	44.9 (1.9)	18.5
	ML	3.40*§	0.30	43.9 (2.1)	24.1
	LG	0.76*	0.13	20.7 (1.3)	41.5
Angular momentum	AP	3.06†	0.22	42.8 (1.3)	2.0
	ML	2.82†	0.28	41.1 (1.4)	2.8
	LG	0.69*	0.13	20.2 (1.5)	27.5

Significant differences between methods in each group are shown by superscripts ($p<0.05$),
*de Leva vs. the other methods,
§ Hanavan vs. the other methods,
† Jensen vs. the other methods,
%Difference = (other mean – de Leva mean)*100/de Leva mean

4.3.4 Pearson coefficients of correlation of the head-neck-trunk MOI values

Table 4.11 shows the Pearson coefficients of correlation of the head-neck-trunk moment of inertia estimations between the de Leva (1996) and the other methods for all three morphological groups. The overall absolute average coefficient of correlation for the MOI values was 0.64±0.27. The obese subjects displayed the best absolute average correlation (0.79). Hanavan's model had the strongest correlation (0.82) with the de Leva (1996) to estimate the segment's MOI values in obese subjects for all the three principal axes. The worst absolute average correlations of the head-neck-trunk's moments of inertia were found for the angular momentum method (0.53), and for the normal group (0.15). Both Jenson (1978) and the angular momentum methods gave negative correlations for the normal BMI group.

Table 4.11. Pearson coefficients of correlation (*p* values) of the head-neck-trunk moment of inertia estimations between de Leva method and the other approaches for the normal, lean, and obese morphological groups about the antero-posterior (AP), medio-lateral (ML), and longitudinal (LG) axes

Method	Axis	Normal	Lean	Obese
Hanavan (1964)	AP	0.90 (0.005)	0.95 (0.000)	0.94 (0.000)
	ML	0.83 (0.022)	0.95 (0.000)	0.91 (0.002)
	LG	0.32 (0.492)	0.83 (0.011)	0.75 (0.034)
Jensen (1978)	AP	−0.45 (0.309)	0.58 (0.132)	0.75 (0.032)
	ML	−0.44 (0.321)	0.55 (0.157)	0.69 (0.059)
	LG	0.13 (0.775)	0.75 (0.033)	0.74 (0.036)
Angular momentum	AP	−0.13 (0.775)	0.85 (0.008)	0.84 (0.009)
	ML	−0.19 (0.682)	0.59 (0.124)	0.67 (0.070)
	LG	0.14 (0.766)	0.57 (0.138)	0.81 (0.016)

Chapter 5

5. DISCUSSION

This chapter deals first with the effect of the modeling methods on the estimated MOI values of the whole body less the feet in individuals of different BMI representing normal, lean, and obese morphological types. These methods were those of de Leva (1996), Hanavan (1964), Jensen (1978), and two new methods based on an inverse pendulum approach. Then, these methods were compared with the one of de Leva (1996) by Pearson coefficients of correlation and discussed. Afterwards, the effect of the modeling methods on the MOI values of the head-neck-trunk in individuals of different morphology will be argued. These methods were those of de Leva (1996), Hanavan (1964), Jensen (1978), and angular momentum. Since to calculate the head-neck-trunk MOI values its mass and COM location are required, the methods to estimate them in different morphological groups will be discussed beforehand. These methods were those of de Leva (1996), Dempster (1955), Hanavan (1964), Jensen (1978), and a new force-plate technique. Next, the behavior of the methods to estimate the head-neck-trunk MOI values with respect to the de Leva (1996) method for each morphological group are discussed. Finally, the limitations of this study are presented and suggestions for the further studies are offered.

5.1 Effect of the modeling methods to estimate the MOI values of the whole body less the feet in individuals of different body morphology

The first objective of this study was to test the effect of the modeling methods on the MOI values of the whole body less the feet in individuals of different BMI representing normal, lean, and obese morphological types. These methods were those of de Leva (1996), Hanavan (1964), Jensen (1978), and for the first time two new personalized in-vivo methods based on inverse dynamics and angular momentum equations.

For subjects with a normal BMI range, the five methods gave similar MOI values and that for all the three axes. Though not statistically different from the others, the photogrammetric method (Jensen, 1978) displayed the highest values. This can be explained in part by the assumption of a uniformed mass density in each elliptical section for all the body segments. According to Hatze (1980) and Reid and Jensen (1990) sectioning of some segments into elliptical discs lead to an overestimation error. Pearsall and Costigan (1999) and Ganley and Powers (2004) tested several methods (Dempster, 1955; Clauser et al., 1969; Zatsiorsky et al. 1990b; dual energy X-ray absorptiometry) on some gait parameters. They reported low variability in joint kinetic values. This lack of effect can be explained in part by subjects having MOI values within the normal BMI range. However, these authors underlined the importance of segment inertial parameters in movements involving large limb accelerations as encountered in running. In a gait study involving children with a normal BMI (18.5 ± 1.8 kg/m^2), Bauer et al. (2007) reported significant differences in joint moments and powers due to segment inertial parameters measured by MRI and those obtained from Jensen's (1986) regression equations derived from photogrammetry. Nonetheless, they concluded that even the greatest differences in kinetic values were relatively small. It appears that the effect of MOI estimation methods on joint kinetics of subjects within the normal BMI range was negligible. However, in these studies only the lower limbs were considered which represent about 40% of the total body mass. The influence of the trunk mass and MOI values was not assessed. Our results on the whole body moments of inertia support these findings for subjects with normal BMI range.

For subjects within the lean BMI group, the de Leva (1996), Hanavan (1964), and Jensen (1978) methods had higher MOI values than the two proposed methods (inverse dynamics and angular momentum). The MOI values calculated by de Leva (1996) method for the lean adults could be too high. Since it is based on the segments' length and body mass alone, de Leva (1996) method is strongly related to the morphology of the individuals from whom the anthropometric data were collected. Thus, this overestimation could be partially attributed to the differences in the mass distribution and segment densities of the subjects in the lean group from those of the young athletic population tested by Zatsiorsky (1990a).

Hanavan's model (1964) represents the body segments as simple geometric shapes where their mass is estimated using regression equations based on cadaver studies. These could lead to higher MOI values of the body less the feet compared to the personalized proposed methods. To evaluate the accuracy of the inertial parameters in children with ectomorphic, mesomorphic, and endomorphic body types, the Jensen's photogrammetric method (1978) was compared to those of a mathematical model (Hatze, 1980) and cadaver studies. Good agreement was observed with the model proposed by Hatze (1980) but quite inconsistencies were noted with those of the cadavers where data were collected form older subjects. However, the accuracy of the photogrammetric method to estimate the MOI values of adults and elderly was not investigated. In these populations, the body composition is more similar to old cadavers than those of children, where some segments' mass is overestimated compared to living individuals (Pearsall et al., 1996). Thus, it can be expected that inertial properties of the adults obtained from the photogrammetric method be higher than those of the personalized methods such as Hatze's model (1980) or inverse dynamics and angular momentum approaches.

Hanavan (1964) and Jensen (1978) methods appeared to be better than de Leva (1996) method to estimate the MOI values of the body less the feet in subjects within lean BMI group. These methods take the segments' geometry into account. The segments' shape fluctuations are considered by the Jensen (1978) method, as well. Therefore, the Hanavan (1964) and Jensen (1978) models provide more personalized MOI values in lean subjects compared to de Leva (1996) method, where only the segments' length and body mass are considered.

Inverse dynamics and angular momentum methods made no assumption on the mass distribution and geometry of the body segments. None of the limitation of the previous studies such as segments' masses as obtained from cadavers (e.g., Hanavan, 1964) is involved in these novel approaches. All the necessary factors (e.g., COM linear and angular velocities, forces and moments) to estimate the individual MOI values are calculated by direct measurement methods and accurate tools. Therefore, these methods appeared to give more personalized MOI estimations of the body less the feet in lean subjects than those of de Leva (1996), Hanavan (1964), and Jensen (1978).

For the group of subjects in the high BMI range (obese), the de Leva (1996), Hanavan (1964), and Jensen (1978) methods provided lower MOI values than the two proposed methods. For the de Leva (1996) method this can be explained in part by the differences in mass distributions, and the relative mass of the body segments of our obese subjects compared to those participating in the Zatsiorsky's study (1990a). The segment-mass/body-mass ratios in people of different morphology were not equal (Zatsiorsky, 2002). For instance, for two subjects with body mass of 100 kg and 50 kg, respectively, the ratio of the total body mass equals 2.0. The corresponding ratio for the mass of the head is only 1.27, while it is 2.30 for the abdomen mass. Therefore, the ratios of segment-mass/body-mass should be used with caution in different morphological populations. Furthermore, Jensen (1978) and Hanavan (1964) used the density and mass profiles of the body segments obtained from cadavers. These inertial properties are substantially different from those of adults and living subjects (Reid and Jensen, 1990; Pearsall et al., 1994) and made more apparent in our obese group. This observation may account why DeVita and Hortobágyi (2003) found differences in gait parameters due to factors related to body composition and mass in their 21 obese (BMI = 42.3±7.7 kg/m^2) and 18 normal (BMI = 22.7±2.9 kg/m^2) subjects. In their study, the magnitude of the segmental masses of the lower extremity, their moments of inertia, and the locations of the mass centers were estimated from a geometrical model (Hanavan, 1964). DeVita and Hortobágyi (2003) reported that joint moments and powers were identical at the hip and knee but were higher by 88% and 61% at the ankle in obese compared to normal BMI range participants walking at the same speed. These differences could be attributed in part to the inaccuracies in the MOI values of the lower extremity obtained from geometrical model of Hanavan (1964). Personalized methods where segment masses and MOI values determined experimentally could lead to smaller errors in biomechanical modeling of subjects with different body types, especially in populations such for whom no anthropometric data are available.

Contrary to lean subjects, it is possible to take into account excessive localized mass with Hanavan's method (1964) when assessing obese individuals especially for the trunk. For example, Achard et al. (2006) modified the trunk parameters proposed by Hanavan (1964) to match the morphology of obese teenagers (BMI = 40±5.2 kg/m^2). They examined the influence of the anthropometric models (Dempster, 1955; Hanavan, 1964) on the calculation of the vertical jump performance (COM height) as well as on the mechanical internal energy expenditure (MIEE) obtained from inverse dynamic calculations. While the vertical jump performance did not differ, the MIEE was 40% higher based on modified Hanavan's geometrical model (1964) than that obtained from Dempster

(1955). These differences were related to the transverse moments of inertia of the trunk, where the values based on the modified Hanavan method were twice those obtained with the Dempster's method. In addition, Myers and Steudel (1985) demonstrated that the effect of changes in limb mass and its distribution can result in significant differences on the energetic cost of running. The anthropometric and geometric models do not take into consideration mass distribution in limbs which can lead to substantial errors on estimation of energetic cost of human locomotion.

Hanavan (1964) and Jensen (1978) methods appeared to provide more personalized MOI values of the body less the feet than de Leva (1996) method for the obese BMI group, since these take the segments' shape into account. Hanavan (1964) method can be adapted to model the extra mass in the abdomen region of obese subjects. Jensen's model also (1978) accounts for the segments' shape fluctuations that are vital to the accuracy of the MOI estimations especially for the trunk segment. Consequently, the Hanavan (1964) and Jensen (1978) models provide better MOI values of the segments compared to de Leva (1996) method. Modeling the additional mass of body and taking into consideration the body segments' shape by personalized in-vivo methods has the advantage of avoiding the oversimplifying assumptions of de Leva (1996) method. However, excess mass to be modeled must be localized with the Hanavan's model (1964). Inverse dynamics and angular momentum methods make no assumption on segments' shape and density.

The mass and profile assumptions applied in the Hanavan (1964) and Jensen (1978) models are not included in the inverse dynamics and angular momentum methods. These new methods seem to give more personalized MOI estimations of the whole body less the feet in obese population in comparison with those of de Leva (1996), Hanavan (1964), and Jensen (1978) methods. Angular momentum method showed less variability than inverse dynamics approach to estimate the MOI values of the body less the feet. This could be attributed to the use of COM velocities rather than accelerations by the angular momentum method. In addition, the inverse dynamics method erroneously estimated the MOI values of the segments due to inaccuracies of joint forces and moments calculated by an inverse dynamic approach. This is owed to neglecting of the mass-acceleration and the MOI-angular acceleration products of the fix body less the oscillating limb. Thus, angular momentum method is preferred to the inverse dynamics to estimate the segments' MOI values and was used in the calculation of the inertial parameters of the head-neck-trunk. This method estimated the body segments' MOI values during self-imposed oscillations in the range of the other approaches (de Leva, 1996; Hanavan 1964, and Jensen, 1978). Therefore, angular momentum method can be applied to estimate the body segments' MOI values.

In summary, all the methods provide similar MOI values for subjects within normal BMI range. For lean and obese morphological groups, Hanavan (1964) and Jensen (1978) models appeared to be better than the de Leva (1996) method. Hanavan's method (1964) could be applied to model the localized mass in the abdomen region of obese subjects, while Jensen's model (1978) accounts for the segments' shape fluctuations. The inverse dynamics and angular momentum methods make no assumption on the mass distribution and geometry of the body segments. These methods are based on direct measurements of the kinematic and kinetic parameters applied to estimate the MOI values of the body. Thus, the proposed methods provide more reasonable personalized in-vivo estimates of the body less the feet MOI values especially in lean and obese body morphologies. Angular momentum showed less variability to estimate the body less the feet MOI values than those of inverse dynamics method. This could be attributed to the use of velocities rather than accelerations of the body COM by angular momentum method. Thus, angular momentum method can be applied to estimate the body segments' MOI values. This method was preferred to inverse dynamics to estimate the segments' MOI values during self-imposed oscillations and used in the calculation of the inertial parameters of the head-neck-trunk.

5.2 Comparison of the modeling methods to estimate the MOI values of the whole body less the feet in individuals of different body morphology

The second objective of the study was to verify if the modeling methods behaved similarly to that of de Leva (1996) method by means of Pearson coefficients of correlation for each morphologic group. Though the overall coefficient of correlation was good, high correlations were found in subjects with a lean BMI. These subjects were characterized by less body fat (Heyward, 2004) than those belonging to the other morphological groups. Consequently, a small change in body geometry within that group was better reflected by a change in moments of inertia. This might also be explained by less variability of the segment densities in lean individuals in comparison with the obese.

For subjects within the normal and obese BMI ranges, Jensen's method (1978) gave the worst correlations in average with those of de Leva (1996). This could be attributed to the segment densities obtained from cadaver studies and uniform mass distribution assumption (Hatze, 1980). Hanavan (1964), and the proposed methods were well correlated with the de Leva (1996). This may be explained in part by the similar trend of these methods with those of the de Leva (1996) to estimate the body less the feet MOI values in normal and obese subjects.

Changes in the MOI values for the AP and ML axes were well correlated with the de Leva (1996). The poor correlations observed for the LG axis can be explained by smaller range in MOI values (Δ=1.15 kg.m^2) compared to the other axes (Δ=10.49 kg.m^2 for the AP and Δ=9.10 kg.m^2 for the ML axes). Generally all methods were well correlated with de Leva (1996) method with the exception of photogrammetric method (Jensen, 1978).

In summary, Hanavan (1964), inverse dynamics, and angular momentum methods had a high correlation (0.83) to estimate the MOI values related to the de Leva method in the three morphological types. Jensen's method (1978) gave the worst correlations in average with those of de Leva (1996). Due to a larger range in MOI values for the AP and ML axes, generally the methods had stronger correlations with the de Leva method in these axes. The new personalized methods are recommended to estimate the MOI values of the whole body less the feet for subjects with normal, lean, and obese BMI.

5.3 Effect of the modeling methods to estimate the mass, COM, and MOI values of head-neck-trunk in individuals of different body morphology

In this section, the effect of the modeling methods on the MOI values of the head-neck-trunk in individuals of different morphology will be discussed. These methods were those of de Leva (1996), Hanavan (1964), Jensen (1978), and angular momentum approach. To calculate the head-neck-trunk MOI values, its mass and COM location are first required. Thus, the effect of the methods to estimate the segment's mass and COM position along their longitudinal axis in different morphological groups will first be verified. These methods were de Leva (1996) and Dempster (1955) anthropometric methods, Hanavan's models (1964), the photogrammetric method (Jensen, 1978), and a new force-plate technique. Their similarity to the de Leva (1996) method for each morphological group is afterwards discussed.

5.3.1 Effect of methods to estimate head-neck-trunk mass and COM position in individuals of different body morphology

Mass and COM position of the head-neck-trunk are required to compute the segment MOI values. Since body proportions differ in various populations, these inertia properties need to be estimated from individualized methods rather than by using predictive approaches (e.g., anthropometric tables). Personalized in-vivo methods (e.g., new force-plate technique) that take into account the segments' contour and mass distribution can provide more accurate information of the inertial parameters.

The average head-neck-trunk mass estimates for the three morphological groups based on Dempster (1955), Hanavan (1964), and Jensen (1978) methods were higher by 8.6% from those obtained from the new force-plate technique while the de Leva (1996) method gave the lowest values. These inconsistencies are now discussed. Cadaver-based studies like those of Dempster (1955) tended to overestimate the mass of the trunk in comparison with living subject investigations, where estimates ranged from 44.2 to 52.4% (Chandler et al., 1975; Clarys and Marfell-Jones, 1986; Clauser et al., 1969; Dempster, 1955). These differences could be attributed to changes in tissue compositions after death and to the preservation techniques used (Pearsall et al., 1996). The head-neck-trunk mass estimations obtained from Dempster (1955) in this study are in agreement with the previous investigations. The only exception was observed for the obese subjects where the average segment mass obtained from the photogrammetric method (Jensen, 1978) was slightly higher by 1.3% compared to that of Dempster (1955).

According to Jensen and Fletcher (1994), Plagenhoef et al. (1983), and Reid (1984), Hanavan (1964) and Jensen (1978) methods tended to give higher values for the mass of the whole trunk in comparison with other current in-vivo findings. Pearsall et al. (1996) reported trunk mass based on photogrammetry to be approximately 3–10% greater than those reported by computed tomography. This difference could be due to segment density values derived from cadaver studies in the Hanavan and Jensen methods. According to Hatze (1980) and Pearsall et al. (1996) segment densities obtained from cadavers are greater than those of the living subjects. Furthermore, the head-neck-trunk mass estimations based on de Leva (1996) could be assumed to be too low for all the morphological groups. It might be assigned to lower tissue density values of the trunk acquired from gamma mass scanning technique (Zatsiorsky, 1990a) in comparison with cadaver-based studies.

The novel force-plate technique provided the segment's mass values in the range of the other methods while no assumption was made on the mass distribution. None of the previous studies' limitations are inherent to this technique. It is based on direct measurements of the segments' mass. Since segment inertial parameters vary with body dimensions, the average percent mass of segments can be misleading (Yeadon and Morlock, 1989). Similar caution was raised by Jensen (1989) in a review of segment parameters changes during growth. Mass proportions of the segments in different morphological groups are not constant. For instance, the relative mass of the abdomen is larger and the relative mass of the head, feet, and hands is smaller in heavier people than in lighter people (Zatsiorsky, 2002). The segment-mass/body-mass ratios can be used only when the subject's data are close to the average values of the corresponding sample. Therefore, the new force-plate technique appears to be effective in calculating personalized in-vivo values of the body segments' mass in population with different body types.

A substantial variation was also noted on the head-neck-trunk's COM position acquired from different methods and that for all the morphological groups. The anthropometric methods of Dempster (1955) and de Leva (1996) tended to give further values of the head-neck-trunk's COM with respect to the segment's proximal endpoint. That was noted for all the three morphological groups. For example, the segment's COM position acquired from de Leva method was 25.5% further compared to the Jensen's method (1978) for the obese subjects. These dissimilarities could arise from differences in mass distribution and morphology of the segment. In a study to investigate geometric and inertial data of the trunk in adult males by means of computed tomography, Erdmann (1997) reported that the trunk's COM location was similar to that of Dempster (1955). However, his subjects were mostly within a normal BMI range. This is in agreement with the results obtained from the new force-plate technique where the head-neck-trunk COM position of the normal BMI subjects was closer to its proximal endpoint only by 5.4 and 2.5% compared to those of de Leva and Dempster, respectively. This may indicate that in subjects within normal BMI range, the differences in the estimations of the trunk COM position between cadaver studies and in-vivo personalized methods are negligible.

Our data of the head-neck-trunk COM position based on the force-plate technique was significantly closer to its proximal endpoint compared to de Leva (1996) for the lean participants. These were further from Hanavan (1964) for the normal group, and Hanavan (1964) and the Jensen (1978) for the obese subjects. These discrepancies could be again attributed to the differences in trunk morphology of our subjects from the population tested by the previous studies. Higher trunk mass values by Hanavan (1964) and Jensen (1978) methods could also be responsible for these dissimilarities. However, the measures based on the new force-plate technique were in the range of the other methods. This might represent the capability of the proposed method to calculate the segment's COM position compared to the other methods, while no assumption is made in segment inertial properties.

Head-neck-trunk COM measures obtained from the novel force-plate technique showed the largest variability (ranged from 58.9 to 66.4% of the segment's length) among the three morphological groups. This may indicate the sensitivity of the proposed technique to calculate the COM positions of the segments in subjects with different body types. The effects of body type on variations in trunk COM position were also reported by Pavol et al. (2002). They reported trunk COM of older adults was located inferior to those of Plagenhoef et al. (1983) by 4.3% and 0.7% of the segment length in men and women, respectively. The new force-plate technique can provide subject-specific segmental COM location. No assumption of mass distribution and segment geometry involved in the previous studies (e.g., Hanavan, 1964; Jensen, 1978) are made in the proposed technique. This method can be applied to individuals belonging to populations for which segment COM data is rare or non-existent. Since it involves direct calculation of the head-neck-trunk COM position, this method appeared to provide more personalized measures of the segmental inertial data in populations of various morphologies and especially lean and obese subjects.

In summary, the calculations of the head-neck-trunk mass and COM position based on the proposed method were in the range of the other methods. The head-neck-trunk mass was higher for Dempster (1955), Hanavan (1964), and Jensen (1978) methods, and lower for de Leva (1996) method for the three morphological groups compared to the new force-plate technique. The novel force-plate technique made no assumption on the mass distribution and is based on direct measurements. The average segment-mass/body-mass ratio obtained from the proposed technique for the three body types was located in the range of the other methods. Thus, the proposed technique provides more reasonable personalized in-vivo estimates of the segmental mass in populations with various morphologies. For the head-neck-trunk COM position, all the methods provided similar results for the normal BMI group. The new force-plate technique can provide subject-specific segmental COM location. This method showed the largest variability of the segment's COM among the three morphological groups while the results were in the range of the other approaches. It is based on direct measurement while the assumptions of the previous studies are not included. The calculations of the head-neck-trunk's mass and COM position based on the proposed method appeared to be sensitive enough to detect differences between different morphological populations compared to the other methods.

The MOI values of the head-neck-trunk segment are directly proportional to its mass and COM position. Mass and COM location of the segment varied substantially in various morphological groups as obtained from different methods. Thus, the methods employed to estimate mass and COM location can be used to explain the differences in the head-neck-trunk's MOI values. The effect of methods on the segment's MOI values is presented in the following section.

Another goal was to asses if the methods to estimate the head-neck-trunk's mass and COM location behaved similarly to those of de Leva (1996) by means of Pearson coefficients of correlation for each morphologic group. Though the overall coefficients of correlation for mass and COM position were good, higher correlations were found in subjects with a lean BMI compared to the other two groups. This could be attributed again to less body fat in the lean group. A small change in the head-neck-trunk morphology within that group was better reflected by a change in the segment's mass and COM values. The Jensen method (1978) gave the worst correlations for the both segment parameters. This could be attributed in part to the use of segment density values derived from cadaver studies (e.g., Dempster, 1955), the values of which were greater than those estimated from living subjects (Pearsall et al., 1996).

Dempster (1955) and Hanavan (1964) methods behaved very similar to the de Leva method to estimate the segment's mass and COM position ($r = 0.99$ and 0.87, respectively) and for all the morphological groups. This could be explained in part by the constant ratios used in these methods to estimate the segment inertial parameters. Thus, Dempster (1955), Hanavan (1964), and de Leva (1996) methods cannot be expected to behave differently in various body morphologies. Generally, changes in the mass and COM values for all the three morphological groups based on the force-plate technique were well correlated with the de Leva (1996), except for the COM positions for lean and obese groups. Small changes in the head-neck-trunk morphology within those groups were better reflected by a change in the segment COM values as determined by the proposed technique. This could be attributed to morphological differences of our subjects from those of Zatsiorsky et al. (1990a) and the lower correlations between them (0.64 for lean and 0.32 for obese subjects).

In summary, Dempster (1955), Hanavan (1964), and force-plate methods had a higher correlation of about 0.96 to estimate the head-neck-trunk mass related to de Leva method (1996). However, the new force-plate technique had lower correlations related to de Leva method for both mass and COM position compared to those of Dempster (1955) and Hanavan (1964). This represents more sensitivity of the proposed technique to calculate the segmental properties compared to the other methods for lean and obese participants.

5.3.2 Effect of the modeling methods on the MOI values of head-neck-trunk in individuals of different body morphology

The present study aimed to test the capability of each method to estimate the head-neck-trunk segment's MOI values. The MOI values based on the angular momentum approach were compared to the other methods in people with normal, lean, and obese BMI range. These methods were those of de Leva (1996), Hanavan (1964), and Jensen (1978) methods.

For subjects with a normal BMI range, though there were differences between Hanavan (1964) and de Leva (1996) methods, all the methods gave similar MOI values of the head-neck-trunk and that for all the three axes. Angular momentum method provided the MOI values close to the other methods. The highest values were obtained from the photogrammetric method (Jensen, 1978). This can be explained in part by the assumption of a uniformed mass density obtained from Dempster (1955). Thus, the Jensen (1978) method might be improved by using density profiles of the living subjects (Wei and Jensen, 1995). Hatze (1980) and Reid and Jensen (1990) also have some reservation on this method. For instance, Hatze (1980) reported that sectioning the abdomino-pelvic segment into elliptical zones leads to an overestimation of 31% of the principal moments of inertia. Hanavan's model (1964) also gave large MOI values of the head-neck-trunk. In this method, segments masses were estimated using regression equations based on cadaver studies (Barter, 1957). Cadaver studies tend to overestimate segments' masses compared to living subject methods (Pearsall et al., 1996). Thus, higher values for the trunk MOI estimations based on Hanavan (1964) method could be expected. Since the four methods gave similar results and the MOI values obtained from angular momentum method are in the range of the other methods, this method is assumed to be as proper as the other approaches in population within the normal BMI range. It is assumed that all the methods have the ability to provide accurate MOI values of the segment with respect to each other.

For subjects within the lean BMI group, the head-neck-trunk's MOI values obtained from the methods used in this study varied substantially. These differences could be assigned to variations in measurement techniques, their assumptions made to estimate the inertial properties, and lack of ability to conform to personalized data. The method proposed by the de Leva (1996) showed the lowest MOI values for all the three axes. While the MOI values about the LG axis obtained from de Leva (1996) method were close to the other methods, they were significantly lower for the AP and ML axes. This could be due to the lower values of the head-neck-trunk mass (12.0%) obtained with this technique compared to the other methods. To account for the range of trunk inertial properties more accurately, more specific anthropometric measures than segment length and body mass must be considered (Forwood, 1985).

Furthermore, modeling body segments as simple geometric shapes by Hanavan (1964) can affect the MOI values of segments with complex contours like trunk (Rao et al., 2006). This was observed for the MOI values of the head-neck-trunk about the LG axis of the lean BMI group, where the segment's width and depth at the caudal endpoint were employed. While for the AP and ML axes Hanavan (1964) gave comparatively similar estimates, the mean MOI values for LG axis was higher by maximum 29.4% from those of the other methods. The high MOI values of the segment obtained from Jensen (1978) method could again be attributed to sectioning of the abdomino-pelvic into elliptical discs that leads to an over estimation of about 8.5% for the predicted mass, 19% for the vertical position of its COM, and about 31% for the principal MOI values (Hatze, 1980). These differences may explain in part why Davis, 1994; Larish et al., 1987; Pate et al., 1992; and Holt et al., 1990 reported that body segment proportions influence motion patterns and joints kinetic during walking and running. The angular momentum method provided the principal moments of inertia in the range of the other methods, where no assumption was made on the head-neck-trunk's geometry and density profile.

Hanavan (1964) method provided the MOI values for the AP and ML axes close to the average obtained from all the methods and with less variability than those of the de Leva (1996) and Jensen (1978) methods. This method accounts for the trunk's shape without overestimating the inertial parameters. Thus, this method seems to be more appropriate to estimate the MOI values of head-neck-trunk for lean subjects than those of de Leva (1996) and Jensen (1978), except for LG axis.

Angular momentum method involves direct calculations of the head-neck-trunk MOI values about the three principal axes through its COM. This method makes no assumption in the segments density and shape and provides reasonable MOI values. The MOI values of the segment were in the range of those obtained from the other methods. Therefore, angular momentum is recommended to estimate the head-neck-trunk MOI values in lean subjects especially while the MOI values are required along the three principal axes.

Our MOI results showed differences in the head-neck-trunk segment's MOI values between the methods for obese subjects. The methods of Jensen (1978) and Hanavan (1964) provided the highest MOI values while the de Leva (1996) method gave the smallest values especially for the LG axis. This might be owing to neglecting the mass distribution of the segment about LG axis on obese people by de Leva (1996) method. However, a high agreement was observed for the MOI values along the AP and ML axes between the de Leva (1996) and angular momentum methods. In addition, the MOI values of the segment are directly proportional to its mass estimations. The mass values of the head-neck-trunk were higher for the Hanavan (1964) and Jensen (1978) models, and lower for the de Leva (1996) method compared to the force-plate technique. Thus, this is expected that Hanavan (1964), Jensen (1978), and de Leva (1996) methods provide the extreme values of the head-neck-trunk's MOI.

Since Hanavan's model (1964) assumes body segments as simple geometric shapes, the shape fluctuations of the obese trunk appear to be neglected. In obese subjects, the extra mass is largely located in the abdomino-pelvic area and the dimensions of this part are extended to the whole lower trunk. Thus, the segment's volume is overestimated by simplification of the trunk shape that can cause larger MOI values of the head-neck-trunk compared to the other techniques. Furthermore, Hanavan (1964) used Barter's regression equations to estimate segment masses which were derived from a cadaver sample that tend to give higher values for the mass of the trunk in comparison with living subject analysis. The photogrammetric method (Jensen, 1978) was also found to overestimate body segments volume by approximately 10% compared to results of an immersion technique (Kaleps, 1984). Thus, one can expect the head-neck-trunk's MOI values based on Hanavan (1964) and Jensen (1978) methods are being higher compared to in-vivo methods like the angular momentum approach where such limitations are not involved.

Modeling the trunk in obese subjects has been applied in some studies to evaluate the effect of localized extra mass on joint moments. For instance, in an investigation on sit-to-stand movement in normal and obese subjects, Sibella et al. (2003) modeled the fat mass of obese subjects as a hemisphere positioned on the abdomen. The MOI of the hemisphere was estimated based on Dempster (1955). They found high hip joint moment and a minimization of knee joint torque for the obese group compared to the normal participants. Sibella et al. (2003), however, did not apply the other methods to estimate inertial properties of the hemisphere as Achard et al., (2006) employed to analyze torques at their obese subjects' lower limbs joints. The belly model introduced to underline the role of the fat mass in obese people does not take into consideration mass distribution in abdomen which can lead to substantial errors on estimation of kinetic analysis during human performance. This emphasizes the application of more personalized methods to calculate the trunk MOI values in which mass distribution is taken into account.

The de Leva (1996) method gave the MOI values of the head-neck-trunk for the AP and ML axes close to the average obtained from all the methods with less variability than those of the Hanavan (1964) and Jensen (1978). This method appeared to take into account the mass distribution of the segment in obese subjects. Thus, this method seems to be more appropriate to estimate the MOI values of head-neck-trunk for obese subjects than those of Hanavan (1964) and Jensen (1978), except for LG axis.

Angular momentum method is a personalized in-vivo method which is not affected by the segment geometry and mass distribution. The limitations of the current studies related to the segments inertial properties are not taken in. To calculate the head-neck-trunk MOI values based on the angular momentum method, all the required kinematic and kinetic parameters can be calculated by means of accurate instruments (video-based system and force-plate). This method is recommended to estimate the three principal MOI values of the head-neck-trunk segment for the obese subjects.

In summary, the head-neck-trunk's MOI values for subjects in normal BMI group were comparatively at the same range. For lean BMI subjects, Hanavan's method (1964) appeared to be better than those of de Leva (1996) and Jensen (1978) to estimate the segment's MOI values about AP and ML axes. For the obese group, de Leva (1996) method seemed to be more suitable to estimate the MOI values about the AP and ML axes than those of Hanavan (1964) and Jensen (1978). Angular momentum method involves direct calculations of the head-neck-trunk MOI values about the three principal axes, while the limitations of the other methods are not involved into the proposed approach. This method has the capability to provide the three principal MOI values in the range of the other methods while it is sensitive enough in population with normal, lean, and obese body morphologies.

5.4 Similarity of the modeling methods to estimate the MOI values of head-neck-trunk in individuals of different body morphology

The last objective of the study was to verify if the methods employed to estimate the head-neck-trunk MOI values behaved similarly to the de Leva (1996) method for each morphologic group. While the coefficients of correlation for both lean and obese groups were good (0.74 and 0.79, respectively), the methods performed differently from the de Leva method in the group of normal BMI range. This could be owing to larger variability of the methods to estimate the segment's MOI values in lean and obese BMI groups. The lean and obese subjects were characterized by distinct body fat than those belonging to the normal group. Therefore, little changes in head-neck-trunk morphology within those groups were better revealed by changes in the MOI values. This could represent more sensitivity of the other methods to the subjects' body morphology than that of de Leva (1996) method.

The similar trend of the Hanavan's model (1964) to the de Leva (1996) method in all the morphological groups could be due to the use of constant ratios to estimate the MOI values. Again, the Jensen's model (1978) provided the worst correlations which could be attributed to its uniform mass distribution assumption (Hatze, 1980) and overestimation of the segment's volume (McConville and Clauser, 1976; Young et al., 1983). The angular momentum method gave also poor correlations with de Leva (1996) method especially for normal and lean groups. This discrepancy of the MOI values may be explained by the fundamental difference in how these values were obtained. de Leva (1996) developed regression equations of Zatsiorsky et al., (1990a) obtained from gamma mass scanning, whereas, in this study direct calculation of the head-neck-trunk's MOI values were performed.

In general, the MOI estimations acquired from Jensen (1978) and angular momentum methods had lower correlation with those of de Leva (1996) compared to Hanavan's model (1964). Jensen's model gave higher values for the head-neck-trunk's MOI values in all the morphological groups owing to density assumptions and overestimation of the segment's volume. Angular momentum method seems to be sensitive enough to determine the MOI values of head-neck-trunk in population with different morphology.

5.5 Limitations of the study

This section deals with the limitations in the interpretation of the results obtained in this study. One limitation is related to the absence of true measures of body segment inertial properties. To solve in part this problem, three commonly used methods (de Leva, 1996; Hanavan, 1964; Jensen, 1978) were selected to serve as a basis for comparison with the proposed techniques (inverse dynamics and angular momentum). Though there were significant differences between the methods and in different morphological groups, the average MOI values of the whole body less the feet and the head-neck-trunk segments obtained by the proposed techniques were in the range of the other methods. Although many factors could account for these differences, they could be due to limitations of the other methods to estimate the segments' MOI values in population with distinct morphologies (Jensen, 1986; Pearsall et al., 1994). It also provides useful supplemental information for interpretation of the novel methods.

The second limitation is related to the approximation of the upper and lower limbs' MOI values. Given the angular momentum equation, the position vectors of the segments' COM to their proximal joint centers and their masses are constant. Thus, the MOI values of the oscillating limbs are directly related to their linear and angular velocities, and angular momentums. The velocities and angular momentums of the limbs during the periods of oscillation varied. For instance, the linear and angular velocities were reaching to the lowest and highest values at the ends and in the middle of the oscillation arc, respectively. In addition, the oscillating limbs' angular momentum at their proximal joint centers could be calculated accurately during the oscillations. Using the average of the limbs' velocities and angular momentum instead of an instantaneous peak value (Bouisset and Pertuzon, 1968) could better support the assumption that the oscillations of the limbs are affected only by their inertia properties. Thus, the average MOI values of the upper and lower limbs reflected truer MOI estimations than instantaneous values because of considering changes in velocities and angular momentum. The average MOI values of the limbs based on the angular momentum equation through their COM were in the range of the other three methods. Despite providing the average values, the angular momentum equation gave reasonable estimates of the upper and lower limbs' MOI values.

The third limitation is related to the generalization of the whole body less the feet and head-neck-trunk COM measurements taken in supine position to the upright position of the body. In supine position the internal organs and mobile visceral contents of the abdominal cavity are displaced. However, the extent to which this displacement changes the mass distribution and COM position of the segment is not known. The segment properties of the trunk change with orientation in space and with the phases of inspiration and expiration (Zatsiorsky, 2002; Pearsall et al., 1996). According to Zatsiorsky et al. (1981), the difference in the longitudinal location of the COM in standing and supine postures does not exceed 1%. Therefore, we can assume that the COM positions of the whole body less the feet and head-neck-trunk segments calculated in supine position do not affect significantly their MOI values.

Another limitation is related to the ignoring the impulse transfers between segments during the oscillations. The relative displacements of the segments with respect to each other and their out of plane motions can affect the MOI estimations. However, the extent to which these parameters change the angular momentum of the segments is not known. Since the relative displacements of the segments and their out of plane deviations were minimal and in an expected range, we can assume the impulse transfers are negligible and do not have a considerable effect on the MOI estimations.

5.6 Future studies

The present study attempted to test the capability of the inverse dynamics and angular momentum methods for evaluating MOI values in people with different body morphology. A new force-plate technique to calculate segment masses and COM locations was also developed and tested in different morphological groups. There is a need, however, to compare these methods with other techniques such as MRI and CT imaging, in order to validate them. At the time of this study, there was difficulty in finding MRI and CT imaging in the same clinical setting. This validity experiment could be possible in the future.

This study indicated the ability of the angular momentum approach to estimate the MOI values of the body segments. Since the methods available for calculating the body segments' MOI values have several limitations, the use of in-vivo personalized methods where no assumption is made (e.g., the proposed methods) appears to be required. To be used in a clinical setting, the inverse dynamics method should also be developed to estimate the body segments' MOI values. Then, its capability to estimate the MOI values needs to be evaluated by comparing to the corresponding values obtained from the other methods in different morphological body types. This development and validity test could be achieved in the near future.

Many studies considered the effects of anthropometric data on joint kinetics during various movements (Rao et al., 2006; Silva and Ambrosio, 2004; Cahouët et al., 2002; Andrews and Mish, 1996; Pearsall and Costigan, 1999; Challis and Kerwin, 1996). However, the influence of segment inertial parameters was a controversial issue especially during gait. The effect of inertial properties on joint kinetics was assessed in subjects within the normal BMI range. Further research is suggested to investigate whether the MOI estimations from in-vivo personalized (e.g., angular momentum) and common methods could yield distinguishable differences in joints kinetics in population with different morphology. Such investigations could lead us either to accept the common method or to use subject-specific measures of inertia properties.

Although postural stability has been the subject of many studies (Winter, 1995; MacKinnon and Winter, 1993; Morasso and Schieppati, 1999; Rietdyk et al., 1999), few have actually investigated the effect of segment inertial parameters on postural control (Kingma et al., 1996). To our knowledge, there is no published research to report the subject-specific inertial parameters effect on joint moments during balance and postural control. To evaluate the control of the whole body balance during quiet standing and walking using inverted pendulum model, the moments of inertia of the whole body, and head-arms-trunk and swing leg are required, respectively. Analyzing the effect of the personalized MOI values of the body, determined by the proposed methods, during balance and postural control throughout standing and walking can provide clinicians more accurate information about postural control mechanisms. This could be especially important to evaluate postural control of subjects with balance abnormalities (e.g., scoliosis) and populations with distinct morphology like obese.

Chapter 6

6. CONCLUSION

This research project investigated the effect of modeling methods on the MOI values of the whole body less the feet and head-neck-trunk segments. Two new personalized in-vivo methods (inverse dynamics and angular momentum approaches) were proposed here to estimate the MOI values in subjects with different morphology and compared to those of de Leva (1996), Hanavan (1964), and Jensen, (1978). Different body types (normal, lean, and obese BMI groups) were involved to test the capability of the methods for estimating the MOI values in various morphological populations.

All methods provided similar MOI values of the whole body less the feet for subjects with a normal BMI range and that for all the three principal axes. For lean and obese morphological groups, de Leva (1996), Hanavan (1964), and Jensen (1978) methods provided the highest and lowest MOI values, respectively. It may indicate the inaccuracy of these methods when applied on subjects with distinctive body types from whom of the original studies. Though the inverse dynamics and angular momentum methods gave the lowest and highest MOI values for lean and obese subjects, respectively, their average was in the range of the other methods. These personalized methods seemed to be capable to estimate the whole body less the feet MOI values in subjects with different morphologies and especially lean and obese subjects.

Generally all methods were well correlated with the de Leva (1996) to estimate the body less the feet MOI values with the exception of Jensen (1978) method. This was attributed in part to the difficulty of that method to take into account body fat content. Since the proposed methods made no assumption on the mass distribution and segments' geometry, these appeared to be appropriate for obtaining MOI values in various morphological populations.

To calculate the head-neck-trunk MOI values, its mass and COM location are required. A novel force-plate technique was developed to calculate the segment's mass and COM position and the results were compared to those of de Leva (1996), Dempster (1955), Hanavan (1964), and Jensen (1978) in different body types. Cadaver-based methods overestimated the head-neck-trunk's mass and COM compared to the other methods and for all the morphological groups. Hanavan (1964) and Jensen (1978) models gave higher values for the mass to some extent, while this was lower for de Leva (1996) compared to the proposed technique. Though there were some significant differences in mass and COM positions, the new force-plate technique gave these values in the range of the other methods. This technique involves a direct measurement of the segment's mass and COM location and was showed to be sensitive to detect differences between diverse morphological populations.

The head-neck-trunk's mass and COM of Dempster (1955), Hanavan (1964), and force-plate methods had a good correlation to those of the de Leva (1996). Jensen method (1978) gave the worst correlations for the both segment parameters due to density assumption. However, the new force-plate technique had lower correlations related to de Leva method for both mass and COM position compared to those of Dempster (1955) and Hanavan (1964). This represents more sensitivity of the proposed technique to calculate the segmental properties compared to the other methods for lean and obese participants.

The head-neck-trunk segment's MOI values for normal BMI subjects were comparatively at the same range. For lean and obese BMI subjects, Hanavan (1964) and de Leva (1996) methods, respectively, appeared to provide suitable MOI estimations of the segment along AP and ML axes. Angular momentum method was the only method that gave reasonable values for the head-neck-trunk's MOI values in the three morphological groups and for all the principal axes. This method involves direct calculations of the segment's MOI values and the limitations of the other methods are not involved in. This method has the capability to provide the three principal MOI values at the range of the other methods, while it is sensitive to identify the differences in populations with normal, lean, and obese body types.

Hanavan's model (1964) behaved similarly to the de Leva (1996) for each morphologic group to estimate the segment's MOI values. Jensen (1978) technique provided the worst correlations which could be attributed to overestimation of the segment's volume. The angular momentum method gave also poor correlations with de Leva (1996) method for the lean and normal groups.

This could represent more sensitivity of the proposed personalized method to the subjects' body morphology than that of the de Leva method.

In summary, the capability of the two new approaches (inverse dynamics and angular momentum) to estimate the whole body less the feet and head-neck-trunk in various morphological groups was obtained. For the whole body less the feet MOI values, the proposed methods are recommended especially for populations within lean and obese BMI range. To estimate the segment inertia parameters of the head-neck-trunk, the proposed force-plate technique and angular momentum approach are also recommended in population with distinct morphologies.

7. REFERENCES

Abdel-Aziz, Y.I., Karara, H.M., 1971. Direct linear transformation from comparator coordinates into object space coordinates in close-range photogrammetry. *Proceedings of the Symposium on Close-Range Photogrammetry*. Falls Church, VA: American Society of Photogrammetry, pp. 1–18.

Achard de Leluardière, F., Hajri, L.N., Lacouture, P., Duboy, J., Frelut, M.L., Peres, G., 2006. Validation and influence of anthropometric and kinematic models of obese teenager in vertical jump performance and mechanical internal energy expenditure. *Gait and Posture* 23, 149–158.

Ackland, T.R., Blanksby, B.A., Bloomfield, J., 1988. Inertial characteristics of adolescent male body segments. *Journal of Biomechanics* 21, 319–328.

Allard, P., Stokes, I.A.F., Blanchi, J.P., 1995. *Three-Dimensional analysis of human movement.* Chapter 5, by Human Kinetics publisher.

Alum, J.H.J., Young, L.R., 1976. The relax oscillation technique for the determination of the moment of inertia of limb segments. *Journal of Biomechanics* 9, 21–25.

Andrews, J.G., Mish, S.P., 1996. Methods for investigating the sensitivity of joint resultants to body segment parameter variations. *Journal of Biomechanics* 29, 651–654.

Baca, A., 1996. Precise determination of anthropometric dimensions by means of image processing methods for estimating human body segment parameter values. *Journal of Biomechanics* 29, 563–567.

Barbier, F., Allard, P., Guelton, K., Colobert, B., Godillon, A., 2003. Estimation of the 3-D center of mass excursion from force plate data during standing. *IEEE Transactions on Neural System and Rehabilitation Engineering* 11, 31–37.

Barter, J.T., 1957. *Estimation of the mass of body segments*. Technical Report 57–260, Wright Patterson Air Force Base, OH.

Bauer, J.J., Pavol, M.J., Snow, C.M., Hayes, W.C., 2007. MRI-derived body segment parameters of children differ from age-based estimates derived using photogrammetry. *Journal of Biomechanics* 40, 2904– 2910.

Behnke, A.R., 1961. Comment on the determination of whole body density and a resume of body composition. In Brozck J. (ed): *Techniques for measuring* *body compositions*. Washington D.C., National Research Council, pp. 118–133.

Bernstein, N.A., Salzgeber, O.A., Pavlenko, P.P., Gurvich, N.A., 1931. *Determination of location of the centers of gravity and mass of the limbs of the living human body* (in Russian). Moscow: All-Union Institute of Experimental Medicine.

Bouisset, S., Pertuzon, E., 1968. Experimental determination of the moments of inertia of limb segments. In Wartenweiler J (ed): *Biomechanics I*, New York, S Karger, pp. 106–109.

Braune, W., Fischer, O., 1889. *The center of gravity of the human body as related to the equipment of the German Infantry*. Abhandlungen der mathematisch-physischen Klasse der Koniglich–Sachsischen Gesellschaft der Wissenschaften 26, 561–672.

Brenière, Y., 1996. Why we walk the way we do. *Journal of Motor Behavior* 28, 291–298.

Brooks, C.B., Jacobs, A.M., 1975. The gamma mass scanning technique for inertial anthropometric measurement. *Medical Science Sports* 7, 290–294.

Cahouët, V., Martin, L., Amarantini, D., 2002. Static optimal estimation of joint accelerations for inverse dynamics problem solution. *Journal of Biomechanics* 35, 1507–1513.

Carpenter, M.G., Frank, J.S., Winter, D.A., Peysar, G.W., 2001. Sampling duration effects on centre of pressure summary measures. *Gait and posture* 13, 35–40.

Casper, R.M., Jacobs, A.M., Kennedy, E.S., McMaster, I.B., 1971. On the use of gamma ray images for determination of body segment parameters. Paper, *presented at Quantitative Imagery in Biomedical Sciences*, Houston, TX.

Challis, J.H., Kerwin, D.G., 1996. Quantification of the uncertainties in resultant joint moments computed in a dynamic activity. *Journal of Sports Sciences* 14, 219–231.

Chandler, R.F., Clauser, C.E., McConville, J.T., Reynolds, H.M., Young, J.W., 1975. *Investigation of inertia properties of the human body.* AMRL Technical Report 74–137, Wright Patterson Air Force Base, OH.

Cheng, C.K., Chen, H.H., Chen, C.S., Lee, C.L., Chen, C.Y., 2000. Segmental inertial properties of Chinese adults determined from magnetic resonance imaging. *Clinical Biomechanics* 15, 559–566.

Clarys, J.P., Martin, A.D., Drinkwater, D.T., 1984. Gross tissue weights in the human body by cadaver dissection. *Human Biology* 56(3), 459–473.

Clarys, J.P., Marfell-Jones, M.J., 1986. Anatomical segmentation in humans and the prediction of segmental masses from intra-segmental anthropometry. *Human Biology* 58, 771–782.

Clauser, C.E., McConville, J.T., Young, J.W., 1969. *Weight, volume and center of mass of segments of the human body.* AMRL Technical Report 69–70. Wright Patterson Air Force Base, OH.

Davis, B.L., 1994. Joint moments: evaluation of ground reaction, inertial and segmental weight effects. *Gait and Posture* 2(1), 58.

De Leva, P., 1996. Adjustments to Zatsiorsky-Seluyanov's segment inertial parameters. *Journal of Biomechanics* 29, 1223–30.

Dempster, W.T., 1955. *Space Requirements of the Seated Operator: Geometrical, Kinematic, and Mechanical Aspects of the Body with special Reference to the Limb*. WADC-TR-55-159. Wright-Patterson Air Force Base, OH.

DeVita, P., Hortobágyi, T., 2003. Obesity is not associated with increased knee joint torque and power during level walking. *Journal of Biomechanics* 36, 1355–1362.

Drillis R., Contini R., Bluestien M., 1964. Body segment parameters. *Artificial limbs* 8, 44–66.

Dumas, R., Aissaoui, R., Mitton, D., Skalli, W., de Guise, J.A., 2005. Personalized body segment parameters from biplanar low-dose radiography. *IEEE Transactions on biomedical engineering* 52(10), 1756–1763.

Durkin, J.L., Dowling, J.J., Andrews, D.M., 2002. The measurement of body segment inertial parameters using dual energy X-ray absorptiometry. *Journal of Biomechanics* 35, 1575–1580.

Durkin, J.L., Dowling, J.J., 2003. Analysis of body segment parameter differences between four human populations and the estimation errors of four popular mathematical models. *Journal of Biomechanical Engineering* 125, 515–522.

Erdmann, W.S., 1997. Geometric and inertial data of the trunk in adult males. *Journal of Biomechanics* 30, 679–688.

Finch, C.A., 1985. *Estimation of body segment parameters of college age females using a mathematical model*. Master thesis, University of Windsor.

Fischer, O., 1906. *Theoretical fundamentals for a mechanics of living bodies, with special applications to man, a well as to some processes of motion in mechanics*. Leipzig: B.G. Tuebner.

Forwood, M.R., Neal, R.J., Wilson, B., 1985. Scaling segmental moments of inertia for individual subjects. *Journal of Biomechanics* 18(10), 755–761.

Ganley, K.J., Powers, C.M., 2004. Determination of lower extremity anthropometric parameters using dual energy X-ray absorptiometry: the influence on net joint moments during gait. *Clinical Biomechanics* 19, 50–56.

Gruen, A., 1997. Fundamentals of videogrammetry–A review. *Human Movement Scince* 16, 155–187.

Hanavan, E. P., 1964. *A mathematical model of the human body*. AMRL Technical Report 64–102. Wright Patterson Air Force Base, OH.

Harless, E., 1860. The static moments of the component masses of the human body. *Treatises of the mathematics-physics class*, Road Bavarian Academy of Sciences. 8: 69–96, 257–294. Trans. FTD Technical Report 61–295. Wright-Patterson Air Force Base, OH, 1962.

Hatze, H., 1975. A new method for the simultaneous measurement of the moment of inertia, the damping coefficient and the location of the center of mass of a body segment. *European Journal of Applied Physiology* 34, 217–226.

Hatze, H., 1980. A mathematical model for the computational determination of parameter values of anthropometric segments. *Journal of Biomechanics* 13, 833–843.

Hatze, H., Baca, A., 1992. Contact-free determination of human body segment parameters by means of videometric image processing of an anthropometric body model. *Proceedings of international SPIE congress on image processing*. San Diego, U.S.A.

Herron, R.E., Cuzzi, J.R., Goulet, D.V., Hugg, J.E., 1974. *Experimental determination of mechanical features of adults and children.* DOT–HS–231–2–397, Washington D.C., U.S., Department of Transportation.

Heyward, V.H., Wagner, D.R., 2004. *Applied body composition assessment*, 2nd ed. by Human Kinetics, Champaign, IL, pp. 76.

Hinrichs, R.N., 1985. Regression equations to predict segmental moments of inertia from anthropometric measurements: An extension of the data of Chandler et al. (1975). *Journal of Biomechanics* 18, 621–624.

Holt, K.G., Hamil, J., Andres, R.O., 1990. The force-driven harmonic oscillator as a model for human locomotion. *Human Movement Science* 9, 55–68.

Huang, H.K., Suarez, F.R., 1983. Evaluation of cross-sectional geometry and mass density distributions of humans and laboratory animals using computerized tomography. *Journal of Biomechanics* 16, 821–832.

Jensen, R.K., 1978. Estimation of the biomechanical properties of three body types using a photogrammetric method. *Journal of Biomechanics* 11, 349–358.

Jensen, R.K., 1986. Body segment mass, radius and radius of gyration proportions of children. *Journal of Biomechanics* 19, 359–368.

Jensen, R.K., Wilson, B.D., 1988. Prediction of segment inertias using curvilinear regression. In Cotton C.E., Lamontagne, M., Robinson, D.G.E., Stothart, J.P.

(eds): *Proceedings of the Fifth Biennial Congress of the Canadian Society for Biomechanics*. Spodym, London, Ontario.

Jensen, R.K., 1989. Changes in segment inertial proportions between four and twenty year. *Journal of Biomechanics* 22, 529–536.

Jensen, R.K., 1993. Human morphology: its role in the mechanics of movement. *Journal of Biomechanics* 26 (Supplement 1), 81–94.

Jensen, R.K., Fletcher, P., 1994. Distribution of mass to segments of elderly males and females. *Journal of Biomechanics* 27(1), 89–96.

Kaleps, I., Clauser, C.E., Young, J.W., Chandler, R.F., Zehner, G.F., McConville, J.T., 1984. Investigation into the mass distribution properties of the human body and its segments. *Ergonomics* 27, 1225–1237.

Kingma, I., Toussaint, H.M., De Looze, M.P., Van Dieen, J.H., 1996. Segment inertial parameter evaluation in two anthropometric models by application of a dynamic linked segment model. *Journal of Biomechanics* 29, 693–704.

Larish, D., Martin, P., Mungiole, M., 1987. Characteristic patterns of gait in the healthy old. *Annals of the New York Academy of Sciences* 515, 18–32.

MacKinnon, C.D., Winter, D.A., 1993. Control of whole body balance in the frontal plane during human walking. *Journal of Biomechanics* 26(6), 633-644.

Martin, P.E., Mungiole, M., Marzke, M.W. Longhill, L.M., 1989. The use of magnetic resonance imaging for measuring segment inertial properties. *Journal of Biomechanics* 22, 367–376.

Matsuo, A. Ozawa, H., Goda, K. Fukunaga, T., 1995. Moment of inertia of whole body using an oscillation table in adolescent boys. *Journal of Biomechanics* 28, 219–223.

McConville, J.T., Clauser, C.E., 1976. Anthropometric assessment of the mass distribution characteristics of the living human body. *Proceedings 6th Congress International Ergonomics Association*, College Park, MD, Human Factors Society, pp. 379–383.

McConville, J.T., Churchill, T.D., Kaleps, I., Clauser, C.E., Cuzzi, J., 1980. *Anthropometric relationships of body and body segments moments of inertia.* Aerospace Medical Research Laboratory Report, AFAMRL–TR–80–119, Wright Patterson Air Force Base, OH.

Monnet, T., Vallée, C., Lacouture, P., 2007. *Journal of Biomechanics (supplement 2)*, pp. S347.

Morasso, P.G., Schieppati, M., 1999. Can muscle stiffness alone stabilize upright standing. *Journal of Neurophysiology* 83, 1622–1626.

Morasso, P.G., Spada, G., Capra, R., 1999. Computing the COM from the COP in postural sway movement. *Human Movement Science* 18, 759–767.

Morlock, M., Yeadon, M.R., 1986. Regression equations for segment inertia parameters. In Allard, P., Gagnon, M., (eds): Human Locomotion IV. Montreal, *Canadian Society for Biomechanics*, pp. 231–232.

Mungiole, M., Martin, P.E., 1990. Estimating segment inertial properties: comparison of magnetic resonance imaging with exiting methods. *Journal of Biomechanics* 23, 1039–1046.

Murray, M.P., Seireg, A., Scholz, R.C., 1967. Center of gravity, center of pressure, and supportive forces during human activities. *Journal of Applied Physiology* 23, 831–838.

Myers, M., Steudel, K., 1985. Effect of limb mass and its distribution on the energy cost of running. *The Journal of experimental biology* 116, 363–373.

Pate, R.R., Macera, C.A., Bailey, S.P., Bartoli, W.P., Powell, K.E., 1992. Physiological, anthropometric, and training correlates of running economy. *Medicine and science in sports and exercise* 24, 1128–1133.

Pavol, M.J., Owings, T.M., Grabiner, M.D., 2002. Body segment inertial parameter estimation for the general population of older adults. *Journal of Biomechanics* 35, 707–712.

Pearsall, D.J., Reid, J.G., Ross, R., 1994. Inertial properties of the human trunk of males determined from magnetic resonance imaging. *Annals of Biomedical Engineering* 22, 692–706.

Pearsall, D.J., Reid, J.G., Livingston, L.A., 1996. Segmental inertial parameters of the human trunk as determined from computed tomography. *Annals of Biomedical Engineering* 24, 198–210.

Pearsall, D.J., Costigan, P.A., 1999. The effect of segment parameter error on gait analysis results. *Gait and Posture* 9, 173–183.

Peyton, A.J., 1986. Determination of the moment of inertia of limb segments by a simple method. *Journal of Biomechanics* 19, 405–410.

Plagenhoef, S., 1971. *Patterns of Human Motion.* Englewood Cliffs, N.J., Prentice-Hall.

Plagenhoef, S., Evans, F.G., Abdelnour, T., 1983. Anatomical data for analysing human motion. *Research quarterly for exercise and sport* 54, 169–178.

Rao, G., Amarantini, D., Berton, E., Favier, D., 2006. Influence of body segments' parameters estimation models on inverse dynamics solutions during gait. *Journal of Biomechanics* 39, 1531–1536.

Reid, J.G., 1984. Physical properties of the human trunk as determined by computed tomography. *Archives of physical medicine and rehabilitation* 65, 246–250.

Reid, J.D., Jensen, R.K., 1990. Human body segment inertia parameters: A survey and status report. *Exercise and sport sciences reviews* 18, 225–241.

Reuleaux, F., 1876. *The kinematics of machinery: outlines of a theory of machines.* London: Macmillan.

Rietdyk, S., Patla, A.E., Winter, D.A., Ishac, M.G., Little, C.E., 1999. Balance recovery from medio-lateral perturbations of the upper body during standing. *Journal of Biomechanics* 32, 1149-1158.

Robertson, D.G.E., Caldwell, G.E., Hamill, J., Kamen, G., Whittlesey, S., 2004. *Research methods in Biomechanics*. Human Kinetics, Champaign, IL.

Sarfaty, O., Ladin, Z., 1993. A video based system for the estimation of the inertial properties of body segments. *Journal of Biomechanics* 26, 1011–1016.

Sibella, F., Galli, M., Romei, M., Montesano, A., Crivellini, M., 2003. Biomechanical analysis of sit-to-stand movement in normal and obese subjects. *Clinical Biomechanics* 18, 745–750.

Silva, M.P., Ambrosio, J.A., 2004. Sensitivity of the results produced by the inverse dynamic analysis of a human stride to perturbed input data. *Gait and Posture* 19, 35–49.

Simmons, J.C., Gardner M.S., 1960. *Self-maneuvering for the orbital worker*. Technical Report TR–60–748. Wright-Patterson Air Force Base, OH.

Smith, J.W., 1957. The forces operating at the human ankle during standing. *Journal of anatomy* 91, 545–564.

Tupling, S.J., Pierrynowski, M.R., Forsyth, R.D., 1984. Anthropometric estimates of the human body using photogrammetry. In *Thornton-Trump A.B. (ed): Human Locomotion III.* Winnipeg, Canadian Society for Biomechanics.

Vaughan, C.L., Davis, B.L., O'Connor, J.C., 1992. *Dynamics of human gait.* Champaign, IL: Human Kinetics.

Wei, C., Jensen, R.K., 1995. The application of segment axial density profiles to a human body inertia model. *Journal of Biomechanics* 28, 103–108.

Weinbach, A.P., 1938. Contour maps, center of gravity, moment of inertia and surface area of the human body. *Human Biology* 10, 356–371.

Whitsett, C.E., 1963. *Some dynamics response characteristics of weightless man.* Master thesis, U.S. Air Force Institute of Technology, Wright-Patterson Air Force Base, OH.

Winter, D.A., 1995. Human balance and postural control during standing and walking. *Gait and Posture* 3, 193–214.

Winter, D.A., 2005. *Biomechanics and motor control of human movement*. Third edition, by John Wiley & Sons Inc.

Yeadon, M.R., Morlock, M., 1989. The appropriate use of regression equations for the estimation of segmental inertia parameters. *Journal of Biomechanics* 22(6/7), 683-689.

Young, J.W., Chandler, R.F., Snow, C.C., Robinette, K.M., Zehner, G.F., Lofberg, M.S., 1983. *Anthropometric and mass distribution characteristics of the adult female*. Technical Report, Oklahoma City, OK, FAA Civil Aeromedical Institute.

Zatsiorsky, V.M., Aruin, A.S., Seluyanov, V.N., 1981. *Biomechanics of masculo-skeletal system*. Moscow: FiS.

Zatsiorsky, V.M., Seluyanov, V.N., 1983. The mass and inertial characteristics of the main segments of the human body. In *Biomechanics VIII-B* (Edited by Matsui, H. and Kobayashi, K.), Human kinetics, Champagne, IL, pp. 1152–1159.

Zatsiorsky, V.M., Seluyanov, V.N., 1985. Estimation of the mass and inertial characteristics of the human body by means of the best predictive regression equations. In *Biomechanics IX-B* (Edited by Winter, D.A, Norman, R.W., Wells, R.P., Hayes K.C., Patla, A.E.), Human kinetics, Champagne, IL, pp. 233–239.

Zatsiorsky, V.M., Seluyanov, V.N., Chugunova, L.G., 1990a. Methods of determining mass-inertial characteristics of human body segments. In *contemporary problems of biomechanics* (Edited by Chernyi, G.G. and Regirer, S.A.), CRC Press, Massachusetts, pp. 272–291.

Zatsiorsky V.M., Seluyanov, V.N., Chugunova, L.G., 1990b. In vivo body segment inertia parameters determination using a gamma-scanner method. In *Biomechanics of human movement: Applications in rehabilitation, sports and ergonomics* (Edited by Berme, N., and Cappozzo, A.) Worthington: Bertec, Ohio, pp. 186–202.

Zatsiorsky, V.M., 2002. *Kinetics of Human Motion.* Champaign, IL: by Human Kinetics.

Zhu, X.P., Checkley, D.R., Hickey, D.S., Isherwood, I., 1986. Accuracy of area measurements made from MRI images compared with computed tomography. *Journal of computer assisted tomography* 10, 96-102.

Appendix A

Research approval by Sainte-Justine Hospital Ethics Committee

LE COMITÉ D'ÉTHIQUE DE LA RECHERCHE

Un comité du CHU Sainte-Justine formé des membres suivants:

Jean-Marie Therrien, éthicien et président

Marie Saint-Jacques, infirmière de recherche

Geneviève Cardinal, juriste

CHU Sainte-Justine
Le centre hospitalier universitaire mère-enfant

Pour l'amour des enfants

Université de Montréal

Les membres du comité d'éthique de la recherche ont étudié le projet de recherche clinique intitulé:

Développement d'une méthode clinique pour estimer le moment d'inertie personnalisé du tronc scoliotique.

No. de protocole: 2341

soumis par: *Paul Allard Ph. D., investigateur principal, collaborateurs: Charles-Hilaire Rivard, M.D, CHU Sainte-Justine, et Mohsen Damavandi, Kinésiologie, Université de Montréal*

et l'ont trouvé conforme aux normes établies par le comité d'éthique de la recherche du CHU Sainte-Justine. Le projet est donc réapprouvé par le Comité.

[redacted]

Président du Comité d'éthique de la recherche

Date de réapprobation: 10 septembre 2007

Appendix B

Mass and COM position of the upper and lower limbs and their sub-segments

Table B.1. Means, standard deviations (S.D.), and the ratios (S.D.) of upper arm mass (kg) as a percent of whole body mass (WBM) along with the % difference between the de Leva (1996) method and the other method. Data are presented for lean (L), normal weight (N), and obese (O) subjects

Method	Group	Mean	S.D.	%WBM	% difference
de Leva (1996)	L	1.41	0.24	2.6 (0.1)	–
	N	2.04	0.08	2.7 (0.0)	–
	O	2.71	0.12	2.7 (0.0)	–
Dempster (1955)	L	1.49*	0.21	2.8 (0.0)	6.03
	N	2.11*	0.08	2.8 (0.0)	3.21
	O	2.80*	0.12	2.8 (0.0)	3.37
Hanavan (1964)	L	0.69*‡	0.31	1.2 (0.4)	−51.27
	N	1.56*‡	0.11	2.1 (0.1)	−23.45
	O	2.55*‡	0.17	2.6 (0.1)	−5.86
Jensen (1978)	L	1.41§	0.27	2.6 (0.3)	−0.05
	N	1.82	0.24	2.4 (0.3)	−10.75
	O	2.45	0.30	2.5 (0.3)	−9.46
Force-plate technique	L	1.55§	0.32	2.9 (0.3)	9.93
	N	1.98‡§	0.13	2.6 (0.1)	−3.00
	O	2.59‡	0.19	2.6 (0.1)	−4.66

Significant differences between methods in each group are shown by superscripts ($p<0.05$),
*de Leva vs. the other methods,
‡Dempster vs. the other methods,
§ Hanavan vs. the other methods,
%Difference = (other mean – de Leva mean)*100/de Leva mean

Table B.2. Means and standard deviations (S.D.) of upper arm center of mass location to the shoulder center (cm), along with the rations (S.D.) as a % of segment length (SL) and the % difference between the de Leva (1996) method and the other methods. Data are presented for lean (L), normal weight (N), and obese (O) subjects

Method	Group	Mean	S.D.	%SL	% difference
de Leva (1996)	L	15.68	1.60	57.6 (0.1)	–
	N	16.36	1.08	57.7 (0.0)	–
	O	16.09	0.57	57.7 (0.0)	–
Dempster (1955)	L	11.86*	1.22	43.6 (0.0)	−24.33
	N	12.36*	0.81	43.6 (0.0)	−24.47
	O	12.15*	0.43	43.6 (0.0)	−24.46
Hanavan (1964)	L	12.91*‡	1.60	47.5 (1.9)	−17.65
	N	13.34*‡	0.99	47.1 (1.3)	−18.46
	O	12.99*‡	0.68	46.6 (1.6)	−19.27
Jensen (1978)	L	13.62*‡	1.23	50.1 (3.0)	−13.13
	N	15.94‡§	0.94	56.2 (2.4)	−2.60
	O	16.35‡§	1.17	58.7 (3.1)	1.62
Force-plate technique	L	15.25‡§	1.38	56.1 (5.1)	−2.71
	N	15.38	2.93	54.2 (10.7)	−6.02
	O	16.17‡§	1.56	58.0 (4.8)	0.52

Significant differences between methods in each group are shown by superscripts ($p<0.05$),
*de Leva vs. the other methods,
‡Dempster vs. the other methods,
§ Hanavan vs. the other methods,
%Difference = (other mean – de Leva mean)*100/de Leva mean

Table B.3. Means, standard deviations (S.D.), and the ratios (S.D.) of forearm and hand mass (kg) as a percent of whole body mass (WBM) along with the % difference between the de Leva (1996) method and the other method. Data are presented for lean (L), normal weight (N), and obese (O) subjects

Method	Group	Mean	S.D.	%WBM	% difference
de Leva (1996)	L	1.12	0.23	2.1 (0.1)	–
	N	1.68	0.06	2.2 (0.0)	–
	O	2.23	0.09	2.2 (0.0)	–
Dempster (1955)	L	1.18	0.17	2.2 (0.0)	4.91
	N	1.66	0.06	2.2 (0.0)	−1.19
	O	2.20	0.09	2.2 (0.0)	−1.34
Hanavan (1964)	L	1.44*‡	0.19	2.7 (0.0)	28.01
	N	1.98*‡	0.07	2.6 (0.0)	18.21
	O	2.60*‡	0.11	2.6 (0.0)	16.53
Jensen (1978)	L	1.34*	0.31	2.5 (0.3)	19.84
	N	1.98	0.38	2.6 (0.4)	17.04
	O	2.57	0.41	2.6 (0.3)	15.19
Force-plate technique	L	1.23§	0.21	2.3 (0.3)	10.79
	N	1.69§	0.12	2.2 (0.1)	0.26
	O	2.29§	0.26	2.3 (0.2)	2.86

Significant differences between methods in each group are shown by superscripts ($p<0.05$),
*de Leva vs. the other methods,
‡Dempster vs. the other methods,
§ Hanavan vs. the other methods,
%Difference = (other mean – de Leva mean)*100/de Leva mean

Table B.4. Means and standard deviations (S.D.) of forearm and hand center of mass location to the elbow center (cm), along with the rations (S.D.) as a % of segment length (SL) and the % difference between the de Leva (1996) method and the other methods. Data are presented for lean (L), normal weight (N), and obese (O) subjects

Method	Group	Mean	S.D.	%SL	% difference
de Leva (1996)	L	16.66	1.44	68.4 (0.6)	–
	N	17.47	0.37	68.4 (0.4)	–
	O	17.24	0.21	68.3 (0.6)	–
Dempster (1955)	L	16.61	1.58	68.2 (0.0)	−0.29
	N	17.41	0.34	68.2 (0.0)	−0.34
	O	17.22	0.26	68.2 (0.0)	−0.09
Hanavan (1964)	L	18.49*‡	1.16	75.9 (2.6)	11.02
	N	17.98*‡	0.14	70.4 (1.1)	2.94
	O	17.08	0.15	67.6 (0.9)	−0.92
Jensen (1978)	L	17.61	2.30	72.3 (6.1)	5.73
	N	20.42	1.96	79.9 (7.9)	16.88
	O	18.36	2.08	72.7 (7.6)	6.53
Force-plate technique	L	16.57§	0.53	68.0 (6.1)	−0.53
	N	17.85	1.90	69.9 (5.6)	2.20
	O	18.17*‡§	0.57	71.9 (2.3)	5.40

Significant differences between methods in each group are shown by superscripts ($p<0.05$),
*de Leva vs. the other methods,
‡Dempster vs. the other methods,
§ Hanavan vs. the other methods,
%Difference = (other mean – de Leva mean)*100/de Leva mean

Table B.5. Means, standard deviations (S.D.), and the ratios (S.D.) of total arm mass (kg) as a percent of whole body mass (WBM) along with the % difference between the de Leva (1996) method and the other method. Data are presented for lean (L), normal weight (N), and obese (O) subjects

Method	Group	Mean	S.D.	%WBM	% difference
de Leva (1996)	L	2.55	0.45	4.7 (0.2)	–
	N	3.72	0.14	4.9 (0.0)	–
	O	4.94	0.21	4.9 (0.0)	–
Dempster (1955)	L	2.67	0.39	5.0 (0.0)	4.95
	N	3.77	0.14	5.0 (0.0)	1.34
	O	5.00*	0.21	5.0 (0.0)	1.29
Hanavan (1964)	L	2.12*‡	0.50	3.9 (0.4)	−16.64
	N	3.55*‡	0.18	4.7 (0.1)	−4.67
	O	5.15*‡	0.28	5.1 (0.1)	4.28
Jensen (1978)	L	2.69§	0.56	5.1 (0.6)	8.75
	N	3.80	0.55	5.0 (0.6)	1.79
	O	5.04	0.60	5.0 (0.5)	1.67
Force-plate technique	L	2.78§	0.49	5.2 (0.4)	10.31
	N	3.66	0.32	4.9 (0.3)	−1.62
	O	4.87§	0.32	4.9 (0.1)	−1.27

Significant differences between methods in each group are shown by superscripts ($p<0.05$),
*de Leva vs. the other methods,
‡Dempster vs. the other methods,
§ Hanavan vs. the other methods,
%Difference = (other mean – de Leva mean)*100/de Leva mean

Table B.6. Means and standard deviations (S.D.) of total arm center of mass location to the shoulder center (cm), along with the rations (S.D.) as a % of segment length (SL) and the % difference between the de Leva (1996) method and the other methods. Data are presented for lean (L), normal weight (N), and obese (O) subjects

Method	Group	Mean	S.D.	%SL	% difference
de Leva (1996)	L	28.06	2.78	54.4 (0.6)	–
	N	29.60	1.57	54.9 (0.6)	–
	O	29.21	0.67	54.8 (0.3)	–
Dempster (1955)	L	27.32*	2.55	53.0 (0.0)	−2.65
	N	28.55*	1.21	53.0 (0.0)	−3.54
	O	28.22*	0.62	53.0 (0.0)	−3.36
Hanavan (1964)	L	35.85*‡	2.27	69.6 (5.5)	27.75
	N	31.79*‡	1.44	59.0 (0.9)	7.40
	O	29.16	0.76	54.7 (0.8)	−0.15
Jensen (1978)	L	30.46*‡§	2.72	59.1 (3.1)	8.53
	N	33.04*‡	1.79	61.3 (2.8)	11.61
	O	31.90‡§	1.90	59.9 (1.5)	9.22
Force-plate technique	L	30.28‡§	3.85	58.7 (3.7)	7.91
	N	30.33‡	1.10	56.3 (2.2)	2.48
	O	29.92‡	1.26	56.2 (1.8)	2.43

Significant differences between methods in each group are shown by superscripts ($p<0.05$),
*de Leva vs. the other methods,
‡Dempster vs. the other methods,
§ Hanavan vs. the other methods,
%Difference = (other mean – de Leva mean)*100/de Leva mean

Table B.7. Means, standard deviations (S.D.), and the ratios (S.D.) of thigh mass (kg) as a percent of whole body mass (WBM) along with the % difference between the de Leva (1996) method and the other method. Data are presented for lean (L), normal weight (N), and obese (O) subjects

Method	Group	Mean	S.D.	%WBM	% difference
de Leva (1996)	L	7.71	1.00	14.5 (0.3)	–
	N	10.67	0.40	14.2 (0.0)	–
	O	14.16	0.60	14.2 (0.0)	–
Dempster (1955)	L	5.34*	0.77	10.0 (0.0)	−30.75
	N	7.54*	0.28	10.0 (0.0)	−29.37
	O	10.00*	0.43	10.0 (0.0)	−29.37
Hanavan (1964)	L	6.41*‡	0.69	12.1 (0.5)	−16.93
	N	8.38*‡	0.26	11.1 (0.1)	−21.44
	O	10.60*‡	0.38	10.6 (0.1)	−25.13
Jensen (1978)	L	6.61*‡	1.25	12.3 (1.1)	−14.74
	N	8.57*	1.40	11.3 (1.6)	−19.97
	O	10.18*	0.53	10.2 (0.5)	−28.04
Force-plate technique	L	5.69*	0.98	10.7 (1.2)	−26.27
	N	8.44*	0.84	11.2 (1.2)	−20.78
	O	11.54*‡§†	0.43	11.5 (0.7)	−18.52

Significant differences between methods in each group are shown by superscripts ($p<0.05$),
*de Leva vs. the other methods,
‡Dempster vs. the other methods,
§ Hanavan vs. the other methods,
† Jensen vs. the other methods,
%Difference = (other mean – de Leva mean)*100/de Leva mean

Table B.8. Means and standard deviations (S.D.) of thigh center of mass location to the hip center (cm), along with the rations (S.D.) as a % of segment length (SL) and the % difference between the de Leva (1996) method and the other methods. Data are presented for lean (L), normal weight (N), and obese (O) subjects

Method	Group	Mean	S.D.	%SL	% difference
de Leva (1996)	L	16.02	2.22	38.5 (2.6)	–
	N	17.02	0.65	40.9 (0.0)	–
	O	17.10	1.08	40.9 (0.0)	–
Dempster (1955)	L	18.00	1.92	43.3 (0.0)	12.36
	N	18.00*	0.69	43.3 (0.0)	5.73
	O	18.08*	1.15	43.3 (0.0)	5.73
Hanavan (1964)	L	16.67	2.17	44.0 (1.0)	4.05
	N	16.48‡	0.91	43.4 (0.8)	−3.20
	O	17.35	1.27	43.4 (1.0)	1.46
Jensen (1978)	L	18.42	2.40	44.3 (6.1)	14.98
	N	19.97*‡§	2.05	48.0 (4.2)	17.29
	O	19.94‡	3.28	47.8 (4.3)	16.64
Force-plate technique	L	16.64	2.35	40.0 (5.1)	3.85
	N	17.76	1.36	42.7 (3.3)	4.30
	O	18.63	4.27	44.6 (10.1)	8.96

Significant differences between methods in each group are shown by superscripts ($p<0.05$),
*de Leva vs. the other methods,
‡Dempster vs. the other methods,
§ Hanavan vs. the other methods,
%Difference = (other mean – de Leva mean)*100/de Leva mean

Table B.9. Means, standard deviations (S.D.), and the ratios (S.D.) of leg and foot mass (kg) as a percent of whole body mass (WBM) along with the % difference between the de Leva (1996) method and the other method. Data are presented for lean (L), normal weight (N), and obese (O) subjects

Method	Group	Mean	S.D.	%WBM	% difference
de Leva (1996)	L	3.14	0.38	5.9 (0.2)	–
	N	4.30	0.16	5.7 (0.0)	–
	O	5.70	0.24	5.7 (0.0)	–
Dempster (1955)	L	3.26	0.47	6.1 (0.0)	3.74
	N	4.60*	0.17	6.1 (0.0)	7.02
	O	6.10*	0.26	6.1 (0.0)	7.00
Hanavan (1964)	L	3.27	0.50	6.1 (0.1)	4.16
	N	4.70*‡	0.18	6.2 (0.0)	9.39
	O	6.30*‡	0.28	6.3 (0.0)	10.53
Jensen (1978)	L	3.26	0.95	6.1 (1.5)	2.58
	N	4.84	0.71	6.4 (0.7)	12.28
	O	5.50§	0.32	5.5 (0.4)	−3.31
Force-plate technique	L	3.45	0.62	6.5 (0.6)	9.26
	N	4.68	0.41	6.2 (0.4)	8.77
	O	5.87*‡§	0.31	5.8 (0.1)	2.87

Significant differences between methods in each group are shown by superscripts ($p<0.05$),
*de Leva vs. the other methods,
‡Dempster vs. the other methods,
§ Hanavan vs. the other methods,
%Difference = (other mean – de Leva mean)*100/de Leva mean

Table B.10. Means and standard deviations (S.D.) of leg and foot center of mass location to the knee center (cm), along with the rations (S.D.) as a % of segment length (SL) and the % difference between the de Leva (1996) method and the other methods. Data are presented for lean (L), normal weight (N), and obese (O) subjects

Method	Group	Mean	S.D.	%SL	% difference
de Leva (1996)	L	23.00	2.51	58.6 (1.1)	–
	N	24.78	1.60	59.9 (0.2)	–
	O	23.63	0.96	60.3 (0.3)	–
Dempster (1955)	L	23.77*	2.61	60.6 (0.0)	3.33
	N	25.06*	1.66	60.6 (0.0)	1.16
	O	23.75	0.98	60.6 (0.0)	0.50
Hanavan (1964)	L	27.27*‡	2.43	69.5 (3.0)	18.59
	N	26.51*‡	1.96	64.1 (0.6)	6.99
	O	23.50	1.32	60.0 (2.3)	−0.53
Jensen (1978)	L	24.03*§	5.83	61.3 (14.5)	4.48
	N	24.48§	1.85	59.2 (3.3)	−1.19
	O	19.72	3.43	50.3 (7.5)	−16.53
Force-plate technique	L	25.54*‡§	2.53	65.1 (2.7)	11.02
	N	24.75	1.81	59.8 (3.9)	−0.09
	O	25.25	1.84	64.4 (3.3)	6.88

Significant differences between methods in each group are shown by superscripts ($p<0.05$),
*de Leva vs. the other methods,
‡Dempster vs. the other methods,
§ Hanavan vs. the other methods,
%Difference = (other mean – de Leva mean)*100/de Leva mean

Table B.11. Means, standard deviations (S.D.), and the ratios (S.D.) of total leg mass (kg) as a percent of whole body mass (WBM) along with the % difference between the de Leva (1996) method and the other method. Data are presented for lean (L), normal weight (N), and obese (O) subjects

Method	Group	Mean	S.D.	%WBM	% difference
de Leva (1996)	L	10.85	1.38	20.4 (0.6)	–
	N	14.97	0.56	19.8 (0.0)	–
	O	19.86	0.84	19.8 (0.0)	–
Dempster (1955)	L	8.60*	1.24	16.1 (0.0)	−20.76
	N	12.13*	0.46	16.1 (0.0)	−18.93
	O	16.10*	0.69	16.1 (0.0)	−18.94
Hanavan (1964)	L	9.68*‡	1.20	18.2 (0.4)	−10.81
	N	13.08*‡	0.44	17.4 (0.1)	−12.60
	O	16.90*‡	0.66	16.9 (0.1)	−14.90
Jensen (1978)	L	9.87	2.02	18.4 (2.0)	−9.72
	N	13.35	2.06	17.7 (2.2)	−10.71
	O	15.69*§	0.65	15.7 (0.7)	−20.94
Force-plate technique	L	9.14*	1.37	17.1 (1.0)	−15.98
	N	13.12*	0.97	17.4 (1.3)	−12.30
	O	17.41*‡†	0.55	17.4 (0.6)	−12.38

Significant differences between methods in each group are shown by superscripts ($p<0.05$),
*de Leva vs. the other methods,
‡Dempster vs. the other methods,
§ Hanavan vs. the other methods,
† Jensen vs. the other methods,
%Difference = (other mean – de Leva mean)*100/de Leva mean

Table B.12. Means and standard deviations (S.D.) of total leg center of mass location to the hip center (cm), along with the rations (S.D.) as a % of segment length (SL) and the % difference between the de Leva (1996) method and the other methods. Data are presented for lean (L), normal weight (N), and obese (O) subjects

Method	Group	Mean	S.D.	%SL	% difference
de Leva (1996)	L	29.92	3.27	37.0 (1.5)	–
	N	31.19	1.11	37.6 (0.7)	–
	O	31.04	1.76	38.3 (0.8)	–
Dempster (1955)	L	36.12*	3.47	44.7 (0.0)	20.71
	N	37.07*	1.55	44.7 (0.0)	18.85
	O	36.18*	0.97	44.7 (0.0)	16.56
Hanavan (1964)	L	33.02*‡	3.44	42.8 (1.5)	10.35
	N	33.74*‡	1.76	42.5 (0.7)	8.17
	O	34.84*	1.79	44.0 (0.9)	12.23
Jensen (1978)	L	33.80*	1.97	41.8 (4.4)	12.96
	N	37.18*§	2.59	44.8 (2.1)	19.20
	O	35.37*	3.92	43.7 (2.8)	13.95
Force-plate technique	L	34.64*	4.66	42.9 (2.6)	15.79
	N	33.90*	2.03	40.9 (2.1)	8.70
	O	35.17*	2.99	43.4 (3.9)	13.32

Significant differences between methods in each group are shown by superscripts ($p<0.05$),
*de Leva vs. the other methods,
‡Dempster vs. the other methods,
§ Hanavan vs. the other methods,
%Difference = (other mean – de Leva mean)*100/de Leva mean