# **Title**: Effect of 3D Printed Foot Orthoses Stiffness on Muscle Activity and Plantar Pressures in Individuals with Flexible Flatfeet: A Statistical nonParametric Mapping Study

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**Word count: Abstract (250); Main text (3759)**

**Running title:** 3D printed foot orthoses in individuals with flatfeet

**Clinical biomechanics:** https://doi.org/10.1016/j.clinbiomech.2021.105553

## ABSTRACT

*Background:* The 3D printing technology allows to produce custom shapes and add functionalities to foot orthoses which offers better options for the treatment of flatfeet. This study aimed to assess the effect of 3D printed foot orthoses stiffness and/or a newly design posting on muscle activity, plantar pressures, and center of pressure displacement in individuals with flatfeet.

*Methods*: Nineteen individuals with flatfeet took part in this study. Two pairs of foot orthoses with different stiffness were designed for each participant and 3D printed. In addition, the flexible foot orthoses could feature an innovative rearfoot posting. Muscle activity, plantar pressures, and center of pressure displacement were recorded during walking.

*Findings*: Walking with foot orthoses did not alter muscle activity time histories. Regarding plantar pressures, the most notable changes were observed in the midfoot area, where peak pressures, mean pressures and contact area increased significantly during walking with foot orthoses. The latter was reinforced by increasing the stiffness. Concerning the center of pressure displacement, foot orthoses shifted the center of pressure forward and medially at early stance. At the end of the stance phase, a transition of the center of pressure in posterior direction was observed during the posting condition. No effect of stiffness was observed on center of pressure displacement.

*Interpretation*: The foot orthoses stiffness and the addition of posting influenced plantar pressures during walking. The foot orthoses stiffness mainly altered the plantar pressures under the midfoot area. However, posting mainly acted on peak and mean pressures under the rearfoot area.

**Keywords: 3D printing, foot orthoses, plantar pressure, muscle activity, flatfoot**

# INTRODUCTION

Flatfoot is defined as a foot deformity that appears or persists after skeletal maturity and is characterized by partial or complete loss of the medial longitudinal arch. The prevalence of flatfeet in the population can be as high as 20–25% (Dunn et al., 2004; Lee et al., 2005). Compared to people with neutral feet, individuals with flatfeet present an altered gait pattern. Thus, it is generally characterized by an excessive rearfoot eversion and a greater forefoot abduction, an increase in the inversion moment at the ankle, as well as an increase in plantar pressures at the medial midfoot area. Foot orthoses (FOs) are commonly prescribed for the treatment of flatfeet and the relief of the pain caused by this condition (Desmyttere et al., 2021). FOs, either custom-made or prefabricated, aim to improve postural stability and lower limb biomechanical parameters during gait such as kinematics, kinetics, electromyography (EMG), and plantar pressures (Banwell et al., 2014; Murley et al., 2009; Redmond et al., 2006). In their meta-analysis, Cheung et al. (Cheung et al., 2011) found that customized FOs are more effective than prefabricated ones in controlling excessive pronation of the foot. However, most of the previous studies have used generic FOs with little adapted to the participants (Aminian et al., 2013; Jafarnezhadgero et al., 2018), thus limiting the application of published research in clinical practice. Moreover, to date only a few studies have investigated the effect of FOs on plantar pressure distribution and muscle activity during gait in individuals with flatfeet.

Regarding plantar pressures, previous studies has provided some evidence that depending on FOs material properties (*e.g.,* density, thickness and ability to adapt to foot contour), pressure distribution may be affected (Gerrard et al., 2020; Hodge et al., 1999; Telfer et al., 2013). It has been shown that softer materials can reduce peak pressures by increasing contact areas during walking (Gerrard et al., 2020). In a similar context, Telfer et al., (Telfer et al., 2013) reported a dose-response effect on plantar pressures when altering the degree of posting inclination. Specifically for flatfeet, FOs aim to compensate for the collapse of the medial longitudinal arch, reduce the forces during heel strike, distribute pressure during the stance phase and facilitate supination for a more efficient propulsion (Desmyttere et al., 2018). A better understanding of the effects of FOs on plantar pressures is essential to enable a targeted treatment specific to flatfeet biomechanics.

Similarly, lower limb muscle activity may also be affected by the wear of FOs. However, contradictory results can be observed in the literature, particularly for the tibialis anterior and the peroneus longus. According to previous studies, FOs can increase (Murley and Bird, 2006), decrease (Dedieu et al., 2013) or have no significant effect (Garbalosa et al., 2015; Telfer et al., 2013) on muscle activity of the peroneus longus. Concerning the tibialis anterior, walking with FOs can increase the activation duration (Tomaro and Burdett, 1993) and the maximum amplitude (Murley and Bird, 2006), having no significant effect (Garbalosa et al., 2015), or decrease the amplitude (Moisan and Cantin, 2016) and the duration of activation (Dedieu et al., 2013). In individuals with flatfeet, Murley et al. (Murley et al., 2010) highlighted a reduction in the activation peak and the root mean square amplitude of the tibialis anterior during loading and conversely, an increase in the activation peak and the root mean square amplitude of peroneus longus between the middle to the end of the stance phase. Other studies assessed the impact of posting on muscle activity, but no significant difference was observed (Murley and Bird, 2006; Telfer et al., 2013). The authors explain the absence of change by the variability present in the response of participants to FOs. In general, several factors such as foot type and FOs designs and material properties might affect these measurements, hence FOs still require further investigation.

Overall, due to the massive variety of FOs geometrical designs, materials and the heterogeneity of protocols that have been used, the current evidence about the effectiveness of FOs on flatfeet is still weak (Desmyttere et al., 2018). The mechanism of action of FOs to improve feet and lower limb function depends on several parameters such as FOs' geometrical design and material properties (Hajizadeh et al., 2020). To the best of our knowledge, no study has investigated the effect of customized FOs design by adjusting the stiffness, on muscle activity and plantar pressure in flatfeet individuals.

The aim of the present study was then to evaluate the effect different levels of stiffness of 3D printed FOs and the addition of an innovative anti-pronator component on muscle activity and plantar pressures in individuals with flexible flatfeet during gait. Our hypothesis was that a lower-level 3D printed FOs stiffness and/or the addition of an innovative anti-pronator component would improve plantar pressure distribution without altering muscle activation patterns in individuals with flatfeet.

## METHODS

### Participants

Nineteen participants with flexible flatfeet (13 Females and 6 Males; age: 37.6±14.0 years; weight: 68.9±11.5 kg; height: 166.7±9.9 m) were included. The inclusion criteria were: (1) a pronated foot type as defined by the foot posture index (≥6) (Redmond et al., 2006), (2) an arch height flexibility >16 mm/kN (Zifchock et al., 2017), (3) no history of wearing FOs prior to this study, (4) no lower limb surgery or injury during the last three months, and (5) having normal lower-limb range of motions and no leg length discrepancy (<0.5 cm) (Burrows, 1966). The participants presented a foot posture index at 7.8±1.3 (range = 6 – 10) and an arch height flexibility equal to 25.6±7.3 mm/kN. All participants gave their written informed consent prior to data collection. This study was approved by the Research Ethics Board of University of Montréal (17–145-CERES-D).

### Experimental procedure

FOs used in this study were customized based on a 3D scan of each participant foot shape, obtained in semi-weight bearing using foot impression boxes while the feet were maintained in a neutral subtalar position by an experienced podiatrist. Detailed description of the FOs design is available in previous studies (Desmyttere et al., 2021, 2020). Briefly, FOs consisted of a plate of 1.5 mm thickness superimposed to honeycombs cells whose height was adjusted as a function of participants' body weight and arch height flexibility to reach two different stiffnesses. Each participant had a 2-weekperiod of familiarization to each FO (flexible and rigid), in a randomized order. After this month of familiarization, the participants were invited to the laboratory to evaluate the effect of the provided FOs on gait parameters. For each participant, the session started with a 5-min of walking familiarization on a treadmill at a comfortable speed which was also useful to determine the speed for the following conditions. Participant were asked to walk as normal as possible in four conditions (control (CO), flexible FOs (F), flexible FOs with posting (P) and rigid FOs (R)). The FOs conditions order was randomized, except for the posting condition which was always performed at the end as the posting was glued on the flexible FOs (see **Fig 1**). The participants were blinded to the conditions. A 5-min rest period was provided between conditions to avoid any fatigue effects.

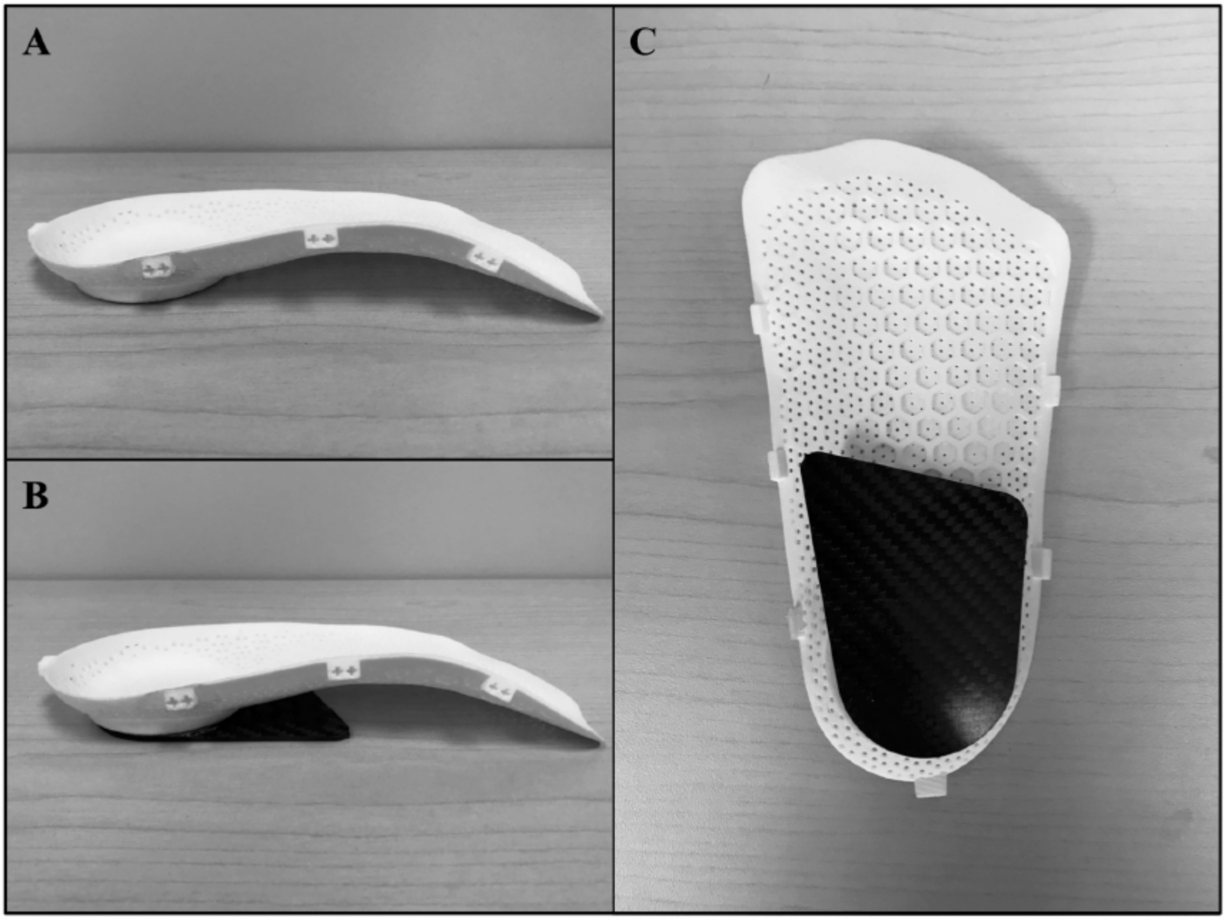

**Fig 1:** Medial view of a flexible FO without (A) and with (B) posting. Bottom view of a flexible FO with posting (C).

### Outcome measures

During the experimentation, neutral running shoes (860 v8, New Balance, USA) were provided to the participants. Four EMG electrodes (Trigno™; Delsys Inc., Boston, MA) placed on the right leg to record the activity of following muscles: Tibialis Anterior (TA), Gastrocnemius Medialis (GM), Soleus (SOL) and Peroneus Longus (PL). Prior to electrode placement, the skin was shaved and cleaned with isopropyl alcohol pads to minimize the impedance between the skin and the electrode. Each electrode location was determined according to the SENIAM recommendations (Hermens et al., 2000). The sampling rate for EMG data was set to 2000 Hz. Moreover, we measured in-shoe plantar pressures as well as the center of pression (CoP) displacement using the Medilogic Flex-Sohle plantar pressure system (T&T Medilogic Medizintechnik GmbH, Germany) at 400 Hz. Plantar pressure insoles were placed between the foot and FO. Although each participant was asked to walk during 3-min in each FO condition, EMG and plantar pressure data were recorded during the last 30-s of each trial to allow participants to familiarize to each condition.

## Data processing

### *EMG data*

Data were analyzed using customized Matlab scripts (Matlab R2021a, The Mathworks, MA). All filters mentioned thereafter are zero-lag 4th-order Butterworth filters. Raw EMGs were band-pass filtered between 10 and 400 Hz. Thereafter, filtered EMG signals were full-wave rectified and envelopes were obtained using a low-pass filter with a cut-off frequency of 9 Hz. Then, the resulting envelopes were normalized by the peak amplitude of the average EMG profile during walking.

### *Plantar pressure insoles data*

For plantar pressure analysis, the foot contact area was divided into seven regions (*i.e.,* medial, and lateral rearfoot, medial and lateral midfoot, and medial, central, and lateral forefoot). The gait events (heel strike and toe off) were detected based on a 10% force threshold (Catalfamo et al., 2008). Peak pressure (N/cm$^2$), mean pressure (N/cm$^2$) and contact area (cm$^2$) were extracted during the stance phase for forefoot, midfoot, rearfoot regions (see **Fig 2**). All data were normalized from 0 to 100% of the stance phase.

**Fig 2.** Representation of the foot contact areas.

## Statistical analysis

One-way ANOVAs using Statistical non-Parametric Mapping (SnPM) (Pataky et al., 2015) were used to assess the effect of FOs stiffness and posting on muscle activity and plantar pressures. In case of significance, post-hoc analysis was performed using paired t-tests. P-values were corrected were adjusted for multiple comparison using Holm-Bonferronni's method (0.05/6 = 0.008) and refined by effect sizes (Cohen's d>0.4). Cohen's d effect sizes were computed over the entire stance phase per post-hoc comparison. If statistical differences were found, only the time periods with a Cohen's d exceeding 0.4 (moderate), for at least 10% of the stance phase, were considered relevant (Armijo-Olivo et al., 2011). When it occurred, the beginning and end of these time periods, and the mean difference (MD) throughout these periods was reported. SnPM analyses were implemented in Matlab R2021a using open-access SPM1D scripts (www.spm1d.org). For muscle activity, a zero-dimensional analysis (ANOVA) was also performed to assess the effect of FOs on the mean muscle activation during the stance phase.

## RESULTS

### *Effect of FOs on muscle activity*

Due to technical problems during data acquisition (*e.g.,* battery and/or signal transmission), some sensors did not provide reliable signals during the tests. Only 17 to 18 out-of-19 participants – with four EMGs available – were included in this analysis (see **Appendix S1**). Averaged patterns of muscle activity are displayed in **Fig 3**. Overall, the ANOVAs (zero-dimensional analysis and SnPM) revealed no significant differences between conditions (control, flexible FO, posting, rigid FO) on muscle activation for all assessed muscles (see **Appendix S2**).

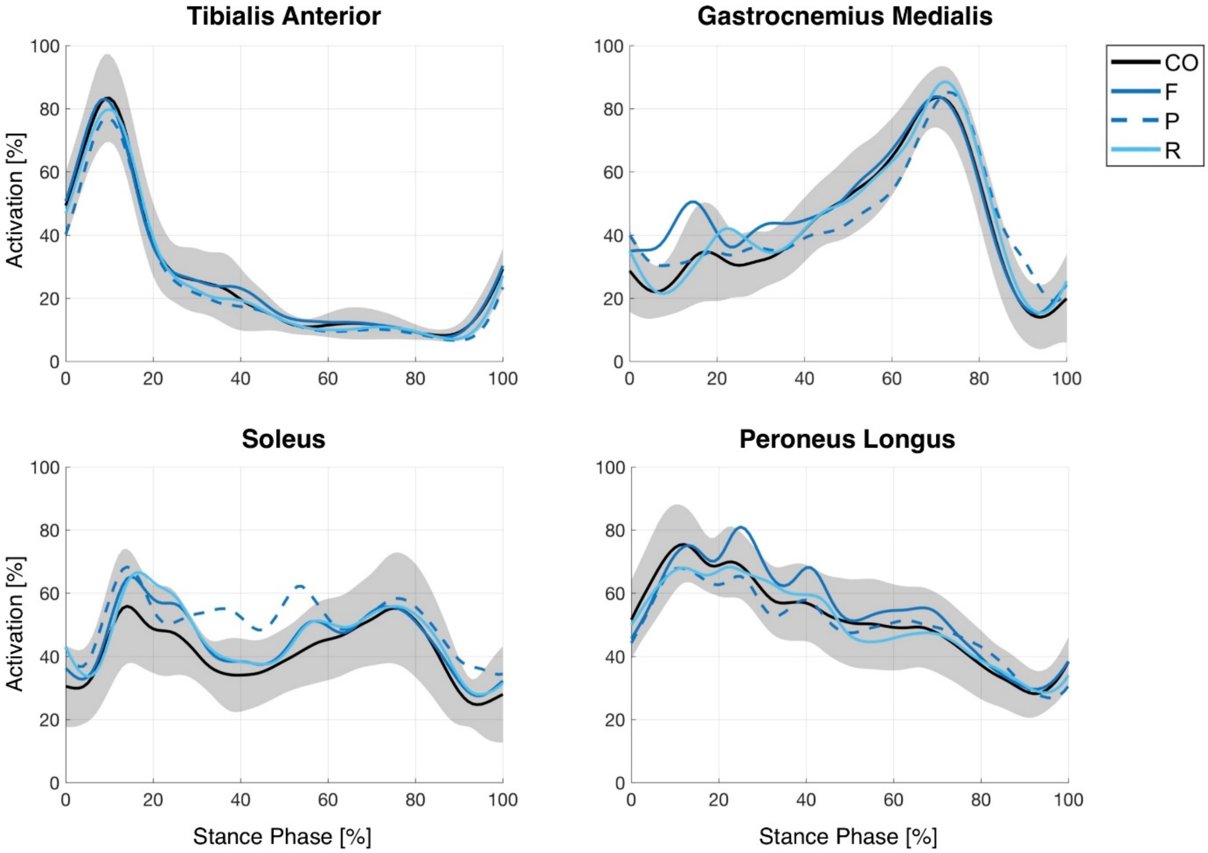

**Fig 3.** EMG patterns for the tested conditions.CO: control, F: flexible FOs, P: flexible FOs with posting, R: rigid FOs.

### *Effect of FOs on plantar pressures*

Concerning plantar pressure changes, ANOVAs indicated significant differences between conditions during the stance phase of walking. Post-hoc analysis results are presented in **Figs 4, 5,**

**6** for the rearfoot, the midfoot and the forefoot areas, respectively. More details concerning plantar pressure statistical results are reported in **Table S3** as supplementary material.

For the **rearfoot area** (see **Fig 4**), a decrease in *peak pressure* was observed while walking with flexible or rigid FOs compared control condition under the medial side (MD = -29.9% – -32.5%, respectively; change observed from 65% to 80% of the stance phase). A significant increase was also observed when walking with FOs compared to the control condition under the lateral side of rearfoot area (MD ranged from +32.0% to +85.0%). However, a significant increase was observed during the posting condition compared to the control condition in the medial (MD = +85.0% change observed from 80% to 100% of the stance phase) and lateral rearfoot areas (MD = +83.0%, change observed from 43% and 100% of the stance phase). The *mean pressure* decreased when wearing the FOs compared to the control condition (MD ranged from -19.1% to -37.8%) during the beginning of the stance phase, under the medial and lateral sides of the rearfoot. However, an increase in mean pressure was observed with the posting compared to control condition at the end of the stance phase (MD = 81.0%). Furthermore, an increase in mean pressure was observed when comparing the posting condition to the flexible and rigid FOs conditions during the second half of stance (MD ranged from +40.9% to +79.5%, respectively). Regarding the *contact area* under the rearfoot area, a decrease was observed when wearing the FOs compared to the control condition (MD ranged from -16.8% to 23.0%; changes observed from 40% to 80% of the stance phase).

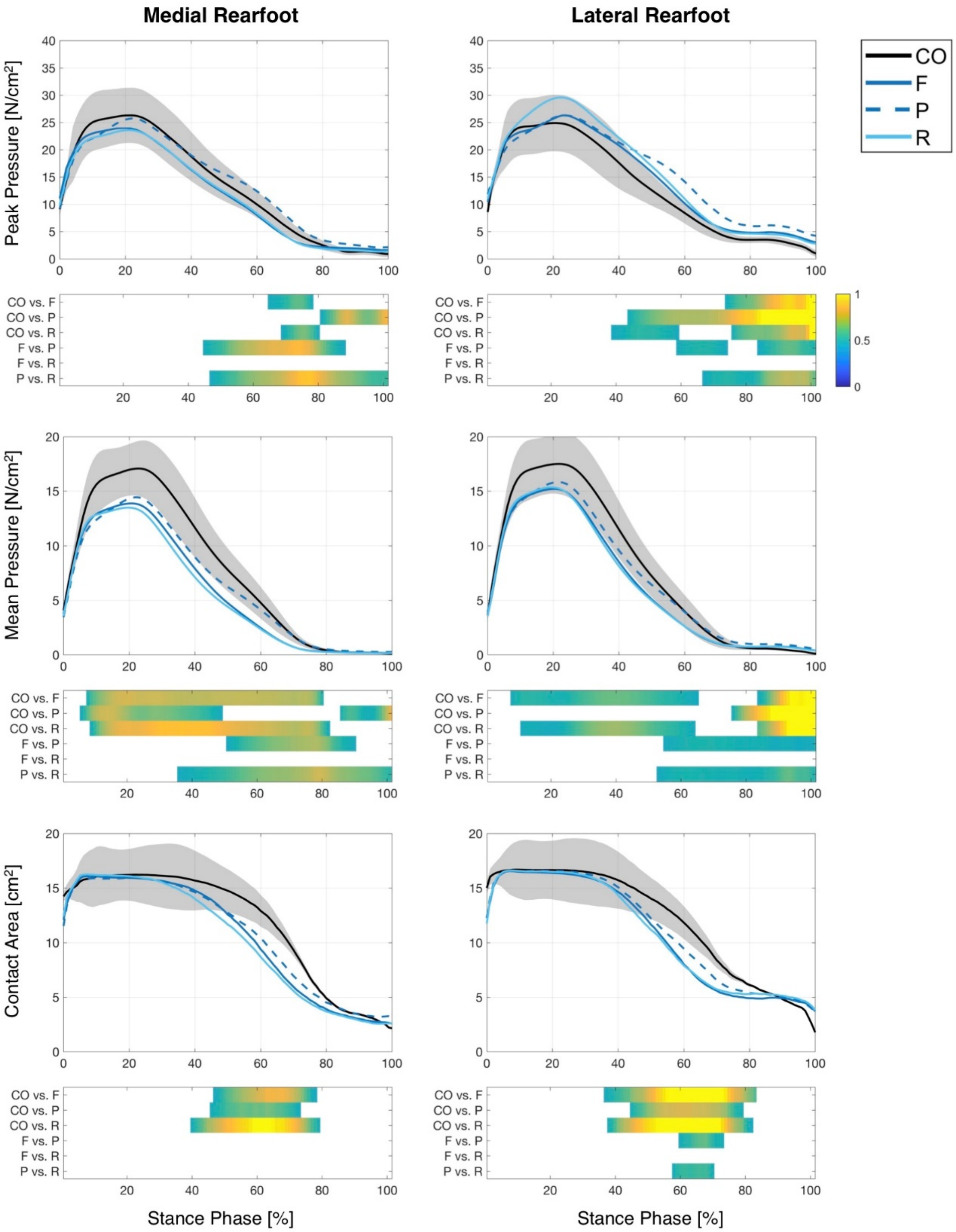

**Fig 4.** Rearfoot peak, mean pressure as well as contact area of the medial and lateral sides. Top graph shows the mean of each condition with 95 % confidence interval cloud (control condition). In the bottom graph, bars indicate significant periods for which the SnPM {t} statistic exceeded the supra-critical threshold ($P < 0.008$). Colormap represents Cohen's d effect size. CO: control, F: flexible FOs, P: flexible FOs with posting, R: rigid FOs.

Concerning the **midfoot area**, the *peak pressure* increased when walking with FOs compared to control condition in both medial and lateral sides (**Fig 5**). A greater increase was observed when wearing the rigid FOs compared to flexible (MD ranged from +13.1% to +23.5%) and posting conditions (MD = +14.3%). Again, an increase was observed in *mean pressure* when wearing the FOs compared to the control condition (MD ranged from +227.1% and +345.0%), throughout the stance phases in both medial and lateral sides. The increase was yet again greater while wearing the rigid FOs compared to posting (MD = 20.7%) and flexible (MD = 34.4%) conditions. Under the lateral side, an increase was also observed during the FOs conditions compared to the control condition (MD ranged between +150.1% and 206.9%), at the end of the stance phase. The *contact area* also increased while wearing the FOs compared to the control condition, regarding the medial side of the midfoot area (MD ranged from +77.7% to +106.1%; changes observed during all the stance phase). A significant increase was observed when comparing contact area using rigid FO comparing to flexible (MD ranged from +11.7% and +19.1% and observed between 35%-45% and 40%-80% of stance phase, respectively) and posting condition (MD = +11.5%; changes observed from 40% to 60%). At the lateral side of the midfoot area, a significant increase at the end of the stance phase was observed during rigid condition (+87.4%) and posting condition (+77.5%) compared to control condition.

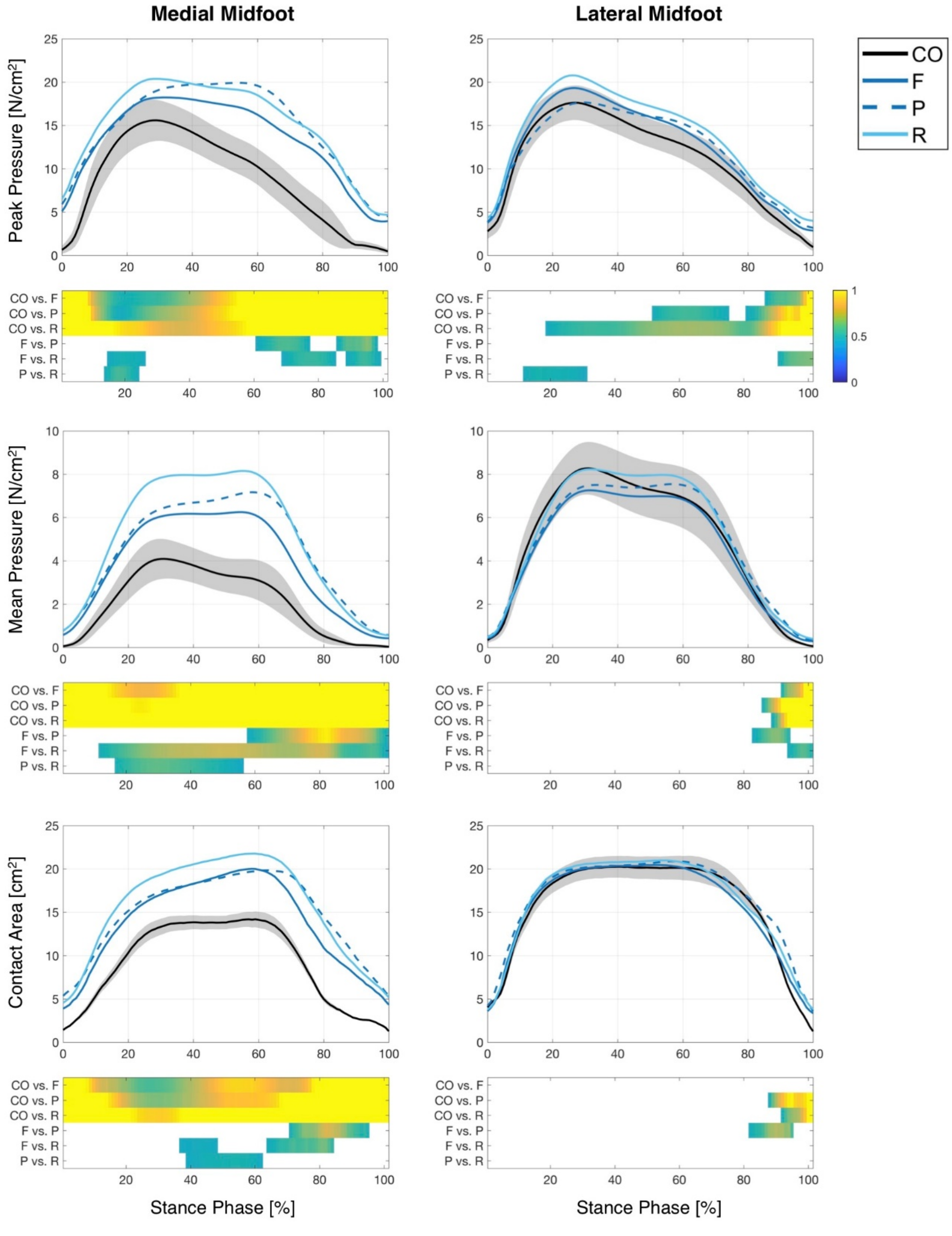

**Fig 5.** Midfoot peak, mean pressure, and contact area of the medial and lateral sides. Top graph shows the mean of each condition with 95 % confidence interval cloud (control condition). In the bottom graph, bars indicate significant periods for which the SnPM {t} statistic exceeded the supra-critical threshold ($P < 0.008$). Colormap represents Cohen's d effect size. CO: control, F: flexible FOs, P: flexible FOs with posting, R: rigid FOs.

Regarding the **forefoot area** (see **Fig 6**), no change was observed in terms of *peak pressure* under the medial side. However, a small decrease was found under the lateral side when walking with rigid or flexible FOs compared to the control condition (MD=-16.6% and -13.9%, respectively). In addition, a brief decrease was observed between posting condition and control condition under the forefoot central area (MD =-12.4%; changes observed from 39% to 58% of stance phase). For the *mean pressure,* a slight decrease was observed under the medial side of the midfoot, only when comparing the posting condition to the control condition (MD=-24.0%; changes observed from 33% to 70% of the stance phase). Under the central side, a decrease in mean pressure was detected during posting condition compared to control condition (MD=-16.7%; changes observed from 37% and 62% of the stance phase) and flexible FOs condition (MD=-14.9%; changes observed from 46% to 56% of the stance phase). Under the lateral side of forefoot area, a little decrease was observed when walking with rigid FO vs control condition (MD=-21.8%; changes observed from 12% to 35% of the stance phase). In term of *contact area*, no significant change was found in the medial and central sides of the forefoot area. However, a small decrease was detected in the lateral side when comparing the rigid FO to the control condition between 0% and 20% of the stance phase (MD=-36.0%).

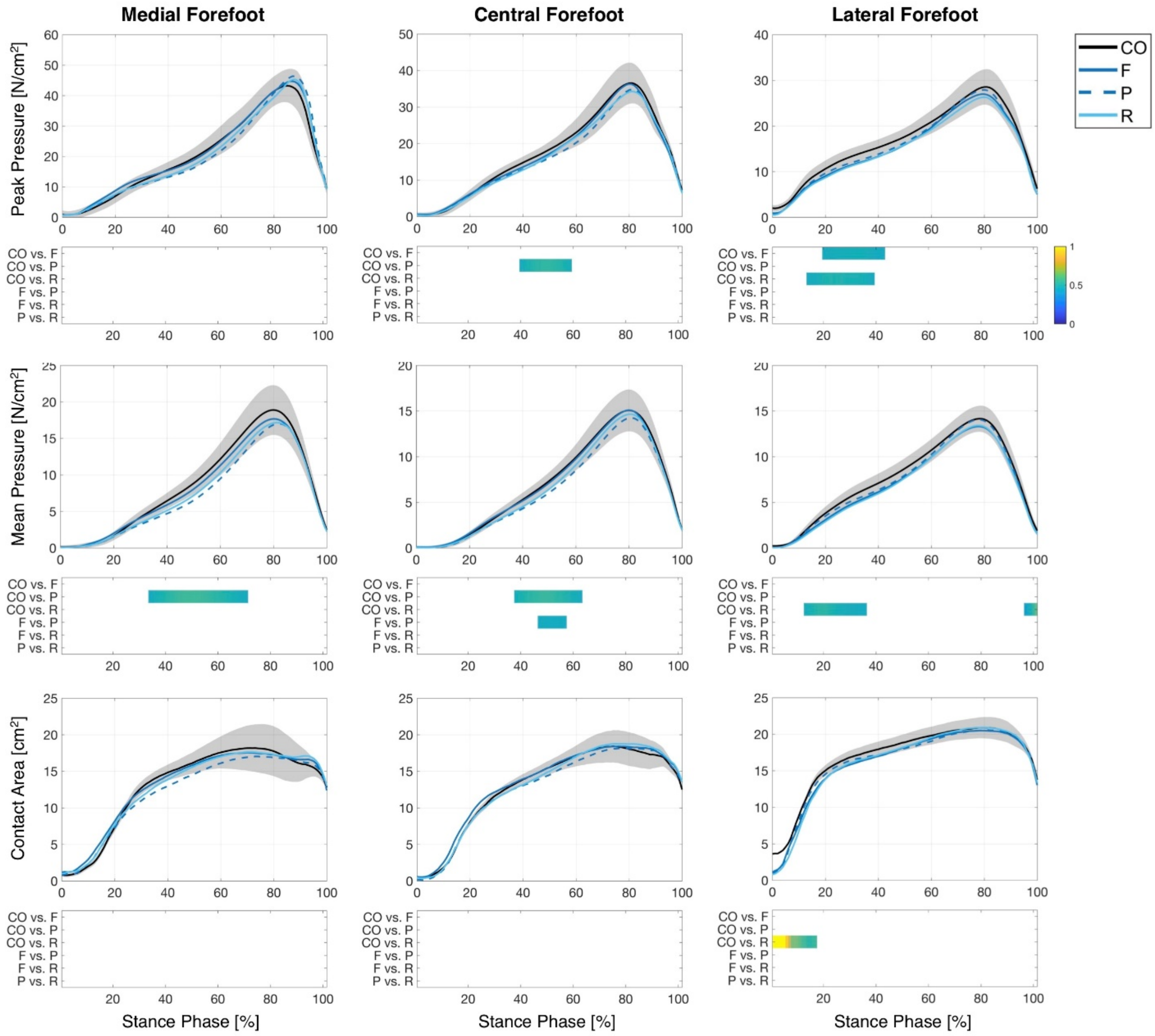

**Fig 6.** Forefoot peak, mean pressure as well as contact area of the medial, central, and lateral sides. Top graph shows the mean of each condition with 95 % confidence interval cloud (control condition). In the bottom graph, bars indicate significant periods for which the SnPM {t} statistic exceeded the supra-critical threshold ($P < 0.008$). Colormap represents Cohen's d effect size. CO: control, F: flexible FOs, P: flexible FOs with posting, R: rigid FOs.

*Effect of FOs on center of pressure displacements*

The anteroposterior and mediolateral displacements of CoP are displayed in **Fig 7**. The CoP patterns in anteroposterior axis showed a difference between conditions. A forward shift has been observed during FOs conditions compared to the control condition (change observed from 0% to 30% of the stance phase). At the end of the stance phase, a transition of the CoP in posterior directions has been observed during the posting condition compared to flexible FOs condition and control condition. Concerning the mediolateral axis, walking with the flexible or rigid FOs lead to a greater medial displacement of the COP compared to the control condition (change observed from 10% to 20% of the stance phase). Posting condition had no effect on CoP mediolateral displacement.

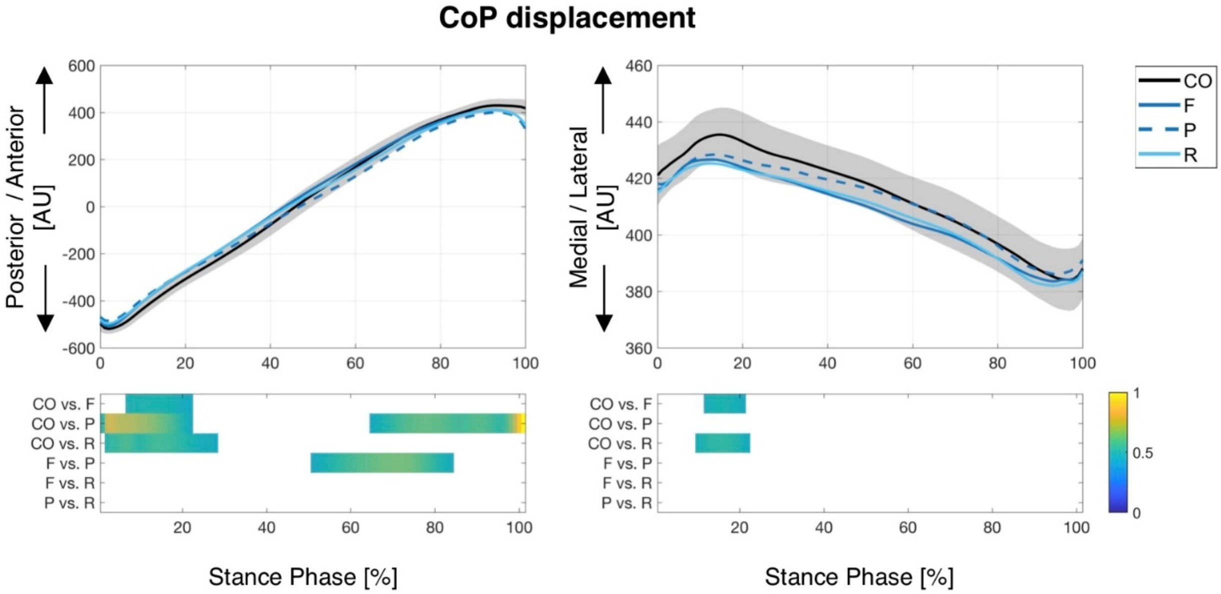

**Fig 7.** Center of pression patterns in the anteroposterior and mediolateral axis. Top graph shows the mean of each condition with 95 % confidence interval cloud (control condition). In the bottom graph, bars indicate significant periods for which a *P*-value < 0.008. Colormap represents Cohen's d effect size. CO: control, F: flexible FOs, P: flexible FOs with posting, R: rigid FOs.

# DISCUSSION

The purpose of this study was to assess the effect of 3D printed FOs stiffness, as well as the addition of an anti-pronator component (i.e., posting) on lower limb muscle activity, plantar pressures, and CoP displacement in individuals with flatfeet. The main findings were that: 1) EMG activity was not altered by wearing FOs and/or the addition of posting, 2) plantar pressures were influenced by the FOs stiffness and the addition of posting, and 3) wearing FOs had little effect on CoP displacements during walking.

### *Effect of FOs on muscle activity*

Contradictory results have been reported by previous studies concerning the effect of FOs on muscle activity. In the present study, no significant change was observed between the different conditions (three FOs conditions and one control condition) in terms of muscle activation profile and amplitude. Those results are in agreement with those reported by Garbalosa et al. (Garbalosa et al., 2015). On the other side, a muscle that was not tested in this study and may be affected by wearing FOs is tibialis posterior (Murley et al., 2010). This muscle acts as a stabilizer of the rearfoot and is particularly affected in the pronated foot type (Barn et al., 2013), however as it is a deep muscle, it cannot be assessed using surface EMG. Furthermore, the absence of changes could be due to the variability present in the response of participants to FOs. In a similar context, Murley & Bird (Murley and Bird, 2006) underlined that the effects of FOs on muscle activation can even be completely opposite between participants and conditions, which is consistent with the paradigm of the "preferred movement pathway" which assumes that the musculoskeletal system has preferential and optimal movement (Nigg et al., 2017, 1999), regardless of the type of movement. Finally, both zero-dimensional and SnPM analyses revealed no significant effect of FOs on muscle activity in individuals with flatfeet. Unlike previous studies that investigated the effect of FOs on muscle activity, a one-dimensional statistical non-parametric mapping analyzes were used in the present study because it takes into account the dependency between time instances of the gait cycle. Using only variables without temporal dimension, the false positive rate can be as high as 76.4% for biomechanical data during locomotion (Pataky et al., 2016).

### *Effect of FOs on plantar pressures*

Regarding plantar pressures, only few studies reported the effect of FOs on plantar pressure distribution in individuals with flatfeet. Looking at individuals with neutral feet, Healy et al. (Healy et al., 2012) reported that the use of denser material for the fabrication of FOs increased average contact area and peak pressures. Recently, Desmyttere et al. (Desmyttere et al., 2020) observed higher peak pressures under the rearfoot and midfoot (up to +31.7 %) when increasing the stiffness of the 3D printed FOs in individuals with neutral feet. Somewhat similar results emerged from the present study in individuals with flatfeet as peak and mean pressures as well as the contact area were increased under the midfoot area (up to +185.0%) when wearing the FOs compared to the control condition. This effect was enhanced when increasing the stiffness. On the other hand, our findings showed a reduction in mean pressures and contact area under the lateral forefoot (up to -21.8% and -36.0%, respectively) and medial rearfoot (up to -32.5 and 27.7%, respectively %) areas when walking with FOs compared to the control condition. Overall, these results are in agreement with those of Redmond et al. (Redmond et al., 2009) who reported a shift of the loads from the rearfoot toward the midfoot  when wearing FOs. Concerning the addition of an anti-pronator component, the impact was more notable at the rearfoot area, where an increase in peak pressures and mean pressures were observed. This impact may be attributed to the carbon stiffness and the heel rise due to the extra thickness (2 mm) induced when using extra-components. In conclusion, the FOs stiffness has the potential to influence plantar pressures as underlined by our findings but also alter foot kinematics as pointed out by our previous studies (Desmyttere et al., 2021, 2020). Further clinical studies on the interaction between stiffness and clinical gait parameters should be carried out in order to find the amount of stiffness that will have an impact on foot kinematic correction without increasing peak pressures.

### *Effect of FOs on CoP displacement*

Regarding the displacement of CoP, it has been noted, in the current literature, that FOs improve postural stability (Masse et al., 2000; Mattacola et al., 2007; Rome and Brown, 2004). In some previous studies, a decrease in the CoP velocity (from 6.5 to 6.8%) (Mattacola et al., 2007) as well as a decrease in the CoP mediolateral displacement (-32%) (Rome and Brown, 2004) have been reported in individuals with flatfeet with the wearing of FOs. Our results underlined a transition in medial and posterior directions toward the beginning of the stance phase, when wearing the rigid or flexible FOs. Thus, a certain improvement of postural stability during walking could be underlined by the reduction of the mediolateral displacement of the CoP when wearing FOs. The cited studies assessed the effect of FOs on the displacement of the CoP during a postural control task, whereas ours measured the CoP displacement during walking.

There are some limitations of this present study that need to be acknowledged. Firstly, there is a methodological limit related to the posting condition which was always carried out at the end as the posting had to be glued on the flexible FOs. Moreover, no familiarization was performed in that condition. Secondly, the posting has only been tested in combination with flexible FOs, while the combination of posting and rigid FOs could provide relevant information. Thirdly, the peroneus longus activity was quantified using surface electrode in the present study, which may explain the absence of difference between conditions.

## CONCLUSION

The 3D printing techniques offer a wide range of possibilities in terms of material properties and design, this study brings new knowledge that could guide the clinical choice to better adapt FOs to individuals with flatfeet, as well as by adding extra-components. Compared to rigid FOs, flexible FOs and posting appeared to be more effective for distributing forces more equally over the plantar surface. Future studies should be performed to find the amount of stiffness that can address the functional needs while avoiding an excessive increase in peak pressures.

## ACKNOWLEDGMENTS

The first author is scholar of Fonds de recherche du Québec – Santé (FRQS). The authors thank the participants who took part of the study. Finally, this project was supported by the TransMedTech Institute and the NSERC R&D Coop with Medicus, Caboma and MedTech.

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**Table S1.** Muscle included in the analysis in function of the subject.

| **subject #** | **TA** | **GM** | **SOL** | **PL** | **total** |
|---|---|---|---|---|---|
| 1 | x | x | x | x | 4 |
| 2 | x | x | x | x | 4 |
| 3 | x | x | x | x | 4 |
| 4 | x | x | x | x | 4 |
| 5 | x | x | x | x | 4 |
| 6 | x | x | x | x | 4 |
| 7 | x | x | x | x | 4 |
| 8 | x | x | x | x | 4 |
| 9 | x | x | x | x | 4 |
| 10 | x | x | x | x | 4 |
| 11 | x | x | x | x | 4 |
| 12 | x | x | Ø | x | 3 |
| 13 | x | x | x | x | 4 |
| 14 | x | x | x | x | 4 |
| 15 | x | x | x | x | 4 |
| 16 | x | x | x | x | 4 |
| 17 | x | x | x | x | 4 |
| 18 | Ø | Ø | Ø | Ø | 0 |
| 19 | x | x | x | x | 4 |
| **total** | 18 | 18 | 17 | 18 | |

The crosses indicate the channels included in the analysis (judged of good quality), and the Ø signs indicate the non-included channels: **TA**: tibialis anterior, **GM**: gastrocnemius medialis, **SOL**: soleus, **PL**: Peroneus Longus.

**Table S2** - Mean muscle activation (%) in each condition during the stance phase of walking

| | **Mean Activation (%) (± SD)** | | | | **ANOVA** |
|---|---|---|---|---|---|
| **Muscle** | **Control** | **Flexible** | **Posting** | **Rigid** | *P* |
| TA | 26.4 ± 5.6 | 26.4 ± 7.5 | 23.5 ± 5.9 | 25.1 ± 6.8 | 0.51 |
| GM | 42.7 ± 8.2 | 46.9 ± 17.2 | 44.1 ± 10.4 | 44.9 ± 11.8 | 0.79 |
| SOL | 41.4 ± 12.6 | 45.2 ± 28.0 | 51.0 ± 45.3 | 47.0 ± 29.4 | 0.84 |
| PL | 52.0 ± 11.8 | 55.1 ± 15.3 | 50.5 ± 15.3 | 51.2 ± 18.7 | 0.83 |

TA: Tibialis Anterior; GM: Gastrocnemius Medial; SOL: Soleus; PL: Peroneus Longus.

**Table S3.** Peak pressure, mean pressure, and contact area results in rearfoot, midfoot, forefoot regions during the stance phase.

| **Foot Region** | **Conditions** | ***Cluster range (% stance)*** | ***Mean difference (%)*** | **Mean effect size** |
|---|---|---|---|---|
| | | **Peak Pressure** | | |
| MR | Control vs. Flex | 64 – 77 | -29.9 | 0.48 |
| | Control vs. Posting | 80 – 100 | +85 | 0.65 |
| | Control vs. Rigid | 68 – 79 | -32.5 | 0.51 |
| | Flex vs. Posting | 44 – 87 | +57.2 | 0.65 |
| | Posting vs. Rigid | 46 – 100 | +37.4 | 0.64 |
| LR | Control vs. Flex | 73 – 100 | +59.9 | 0.77 |
| | Control vs. Posting | 43 – 100 | +83.0 | 0.87 |
| | Control vs. Rigid | 38 – 58 / 75 – 100 | +32.0 / +49.4 | 0.46 / 0.67 |
| | Flex vs. Posting | 58 – 73 / 83 – 100 | +42.2 / +29.8 | 0.45 / 0.51 |
| | Posting vs. Rigid | 66 – 100 | -28.4 | 0.56 |
| MM | Control vs. Flex | 0 – 100 | +140.5 | 1.21 |
| | Control vs. Posting | 0 – 100 | +177.7 | 1.32 |
| | Control vs. Rigid | 0 – 100 | +185.0 | 1.38 |
| | Flex vs. Posting | 60 – 76 / 85 – 97 | +20.3 / +25.2 | 0.48 / 0.53 |
| | Flexible vs. Rigid | 14 – 25 / 68 – 84 / 88 – 98 | +13.1 / +17.2 / +23.5 | 0.43 / 0.47 /0.46 |
| | Posting vs. Rigid | 13 – 23 | +14.3 | 0.47 |
| LM | Control vs. Flex | 86 – 100 | +64.4 | 0.67 |
| | Control vs. Posting | 51 – 74 / 80 – 100 | +17.9 / +65.7 | 0.47 / 0.80 |
| | Control vs. Rigid | 18 – 100 | +38.9 | 0.69 |
| | Flexible vs. Rigid | 90 – 100 | +31.5 | 0.54 |
| | Posting vs. Rigid | 11 – 30 | +20.5 | 0.44 |
| CF | Control vs. Posting | 39 – 58 | -12.4 | 0.46 |
| LF | Control vs. Flex | 19 – 42 | -13.9 | 0.42 |
| | Control vs. Rigid | 13 – 38 | -16.6 | 0.44 |
| | | **Mean Pressure** | | |
| MR | Control vs. Flex | 7 – 79 | -34.4 | 0.69 |
| | Control vs. Posting | 5 – 48 / 85 – 100 | -19.1 / +81.0 | 0.58 / 0.53 |
| | Control vs. Rigid | 8 – 81 | -37.8 | 0.74 |
| | Flex vs. Posting | 50 – 89 | +79.5 | 0.56 |
| | Posting vs. Rigid | 35 – 100 | -40.9 | 0.57 |
| LR | Control vs. Flex | 7 – 64 / 83 – 100 | -23.1 / +91.7 | 0.50 / 0.85 |
| | Control vs. Posting | 75 – 100 | +121.5 | 1.05 |
| | Control vs. Rigid | 10 – 63 / 81 – 100 | -23.7 / +77.8 | 0.52 / 0.86 |
| | Flex vs. Posting | 54 – 100 | +41.4 | 0.44 |
| | Posting vs. Rigid | 52 – 100 | -28.5 | 0.47 |
| MM | Control vs. Flex | 0 – 100 | +227.1 | 1.29 |
| | Control vs. Posting | 0 – 100 | +345.0 | 1.60 |
| | Control vs. Rigid | 0 – 100 | +341.6 | 1.72 |
| | Flex vs. Posting | 57 – 99 | +46.0 | 0.71 |
| | Flexible vs. Rigid | 11 – 99 | +34.4 | 0.64 |
| | Posting vs. Rigid | 16 – 55 | +20.7 | 0.50 |

| | | | | |
|---|---|---|---|---|
| LM | Control vs. Flex | 91 – 100 | +154.7 | 0.85 |
| | Control vs. Posting | 85 – 100 | +150.1 | 1.01 |
| | Control vs. Rigid | 88 – 100 | +206.9 | 1.16 |
| | Flexible vs. Rigid | 82 – 93 | +35.9 | 0.52 |
| | Posting vs. Rigid | 93 – 101 | +40.8 | 0.48 |
| MF | Control vs. Posting | 33 – 70 | -24.0 | 0.47 |
| CF | Control vs. Posting | 37 – 62 | -16.7 | 0.46 |
| | Flex vs. Posting | 46 – 56 | -14.9 | 0.40 |
| LF | Control vs. Rigid | 12 – 35 | -21.8 | 0.44 |
| | | **Contact Area** | | |
| MR | Control vs. Flex | 46 – 77 | -26.1 | 0.65 |
| | Control vs. Posting | 45 – 72 | -18.3 | 0.52 |
| | Control vs. Rigid | 39 – 78 | -27.7 | 0.77 |
| LR | Control vs. Flex | 36 – 84 | -22.5 | 0.87 |
| | Control vs. Posting | 44 – 78 | -16.8 | 0.96 |
| | Control vs. Rigid | 37 – 81 | -23.0 | 0.60 |
| | Flex vs. Posting | 59 – 72 | +19.8 | 0.50 |
| | Posting vs. Rigid | 57 – 69 | -15.4 | 0.47 |
| MM | Control vs. Flex | 0 – 100 | +77.7 | 0.95 |
| | Control vs. Posting | 0 – 100 | +105.3 | 1/18 |
| | Control vs. Rigid | 0 – 100 | +106.1 | 1.35 |
| | Flex vs. Posting | 70 – 94 | +27.1 | 0.60 |
| | Flexible vs. Rigid | 36–47 / 63-85 | +11.7 / +19.1 | 0.41 / 0.51 |
| | Posting vs. Rigid | 38 – 61 | +11.5 | 0.44 |
| LM | Control vs. Posting | 87 – 100 | +77.5 | 0.90 |
| | Control vs. Rigid | 91 – 100 | +87.4 | 0.84 |
| | Flex vs. Posting | 81 – 94 | +25.5 | 0.57 |
| LF | Control vs. Rigid | 1 – 19 | -36.0 | 0.83 |

MR: Medial Rearfoot; LR: Lateral Rearfoot; MM: Medial Midfoot; LM: Lateral Midfoot; LF: Lateral Forefoot