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Abstract: Background. Disruptions to the progress of the centre-of-pressure trajectory beneath prosthetic feet have been reported previously. These disruptions reflect how body weight is transferred over the prosthetic limb and are governed by the compliance of the prosthetic foot device and its ability to simulate ankle function. This study investigated whether using an articulating hydraulic ankle attachment attenuates centre-of-pressure trajectory fluctuations under the prosthetic foot compared to a fixed attachment.
Methods. Twenty active unilateral trans-tibial amputees completed walking trials at their freely-selected, comfortable walking speed using both their habitual foot with either a rigid or elastic articulating attachment and a foot with a hydraulic ankle attachment. Centre-of-pressure displacement and velocity fluctuations beneath the prosthetic foot, prosthetic shank angular velocity during stance, and walking speed were compared between foot conditions.
Findings. Use of the hydraulic device eliminated or reduced the magnitude of posteriorly directed centre-of-pressure displacements, reduced centre-of-pressure velocity variability across single-support, increased mean forwards angular velocity of the shank during early stance, and increased freely chosen comfortable walking speed ($p \leq 0.002$).
Interpretation. The attenuation of centre-of-pressure trajectory fluctuations when using the hydraulic device indicated bodyweight was transferred onto the prosthetic limb in a smoother, less faltering manner which allowed the centre of mass to translate more quickly over the foot.

***Conflict of Interest**

The authors declare that there are no conflicts of interests.

Attenuation of centre-of-pressure trajectory fluctuations under the prosthetic foot when using an articulating hydraulic ankle attachment compared to fixed attachment

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Table 1. Group mean (SD) CoP trajectory measures, shank velocities and walking speeds when walking with *hab*F and *hy*A-F. Participants in Sub-G1 habitually used an Esprit foot and participants in Sub-G2 habitually used a range of other types of feet (see text for detail). Measures that differed significantly when switching to a *hy*A-F are shown in bold. Where differences are significant effect sizes (d) are provided (in italics).

Figure 1. Schematic diagram of hydraulic foot-ankle device (*hy*A-F, Echelon^{TM}). The ankle mechanism allows 6° of plantarflexion (PF) and 3° of dorsiflexion (DF) from the neutral standing position. NB, the part of the foot shown within the cosmesis (greyed-out portion) depicts the type of *hab*F foot (Esprit) habitually used by 12 of the 20 participants.

Figure 2. a) Mean (SD) CoP A-P velocity of the 10 repeat trials, normalised to stance phase, from one participant while using a *hy*A-F (solid line / dark shading) and *hab*F (broken line / light shading). Able-bodied control group CoP velocity ± 1SD ribbon is shown (dotted lines) for reference purposes.

b) Exemplar CoP displacement traces for the same participant shown in figure 2a when using a *hy*A-F (centre) and *hab*F (right). A trace from an able-bodied control is shown for reference purposes (left). All traces are drawn to scale and they have been separated in the medio-lateral direction to allow better view of CoP trajectory fluctuations.

Figure 3. Exemplar mean (SD) shank angular velocity of the 10 repeat trials, normalised to initial double support phase, for one participant when using a *hy*A-F (solid line / dark shading) and *hab*F (broken line / light shading). Able-bodied control group shank angular velocity ± 1SD ribbon is shown (dotted lines) for reference purposes.

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Abstract

Background. Disruptions to the progress of the centre-of-pressure trajectory beneath prosthetic feet have been reported previously. These disruptions reflect how body weight is transferred over the prosthetic limb and are governed by the compliance of the prosthetic foot device and its ability to simulate ankle function. This study investigated whether using an articulating hydraulic ankle attachment attenuates centre-of-pressure trajectory fluctuations under the prosthetic foot compared to a fixed attachment.

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Findings. Use of the hydraulic device eliminated or reduced the magnitude of posteriorly directed centre-of-pressure displacements, reduced centre-of-pressure velocity variability across single-support, increased mean forwards angular velocity of the shank during early stance, and increased freely chosen comfortable walking speed ($p \le 0.002$).

Interpretation. The attenuation of centre-of-pressure trajectory fluctuations when using the hydraulic device indicated bodyweight was transferred onto the prosthetic limb in a smoother, less faltering manner which allowed the centre of mass to translate more quickly over the foot.

INTRODUCTION

During normal able-bodied gait the centre-of-pressure (CoP) progresses throughout stance along the plantar surface of the foot from the heel forwards to the toes. Such progression reflects how the forward progression of the whole body centre of mass is controlled (Schmid et *al*., 2005, Kirtley, 2006). In lower-limb amputees the CoP has been found to remain in the hind-foot area under the prosthetic foot significantly longer than in both the intact or control limbs (Schmid et *al*., 2005), and at times move backwards towards the heel during early-to-mid stance (Ranu, 1988). Anecdotal perceptions of having to 'climb over the prosthetic foot', 'stuttering' or experiencing a 'dead spot' during stance on the prosthetic limb are common features of unilateral amputee gait. Such perceptions are likely to be reflected by interruptions in the forwards progression of the CoP which in turn reflect how bodyweight is transferred over the prosthetic limb (Winter, 2009). In amputee gait CoP forwards progression will be governed by the compliance of the prosthetic foot device (Hafner et *al*., 2002) and in particular its ability to simulate ankle function to provide 1st and 2nd rocker phases of gait.

The functional performance of one particular prosthetic foot versus another is often evaluated using inverse dynamics modelling to determine 'ankle' kinetics for the respective feet. A problem with this approach is that it assumes the foot is a rigid segment with definable 'ankle' joint axes (Winter, 2009). Many current so-called energy-storing and return (ESR) prosthetic feet have no articulating components, and instead deformation of the foot's flexible keels provide simulated dorsi- and plantar- flexion about an undefined axis. These deformations also occur when an articulated connection device is used. Therefore the interpretation of 'ankle' kinetics is at best problematic and sometimes can be misleading (Geil et *al*., 2000, Miller &

Childress, 2005). To avoid such interpretation problems Hansen et *al.,* (2000) proposed using the trajectory of the CoP, transformed from a laboratory-based global coordinate system to the local coordinate system of the shank, to determine the effective 'rocker' or roll-over shape when using a particular prosthetic foot device. In essence the radius and shape of this 'rocker' describes the global functioning of the prosthetic foot-ankle device and removes the necessity of modelling it as a segment and joint. Although this approach has been adopted by others (e.g. Curtze et *al*., 2009, Major et *al*., 2011) a limitation of using roll-over shape characterisation is that it determines the radius of a 'best fit' curve onto a limited number of CoP displacement samples and thus overlooks short-duration disruptions in CoP progression. The magnitude of any such disruptions have been hitherto unmeasured, thus an important characteristic of the prosthetic device is disregarded.

Most current prosthetic feet either have a rigid attachment or incorporate an 'ankle' device allowing elastic articulation. The purpose of the present study was to examine whether use of a foot incorporating a device which allowed hydraulically controlled stance-phase articulation would attenuate the disruptions in CoP progression commonly reported in amputee gait. This foot (Echelon^{TM}, Chas. A. Blatchford and Sons Ltd., Bassingstoke, UK, *hy*A-F) has recently become clinically available and patients who use it report improved comfort and function. When set-up correctly, a *hy*A-F provides 6° plantarflexion and 3° dorsiflexion relative to its neutral (standing) position. We hypothesised that use of a *hy*A-F would facilitate bodyweight transfer onto the prosthetic limb in a smoother less faltering manner, and as a consequence, CoP forward progression would be less disrupted compared to when using participants' habitual feet (*hab*F) with traditional attachment; either non-articulating fixed attachment or elastically controlled articulating device. It was further hypothesised that due to the controlled articulation provided by the *hy*A-F the shank would rotate forwards above the prosthetic foot more 'smoothly' (i.e. with fewer velocity fluctuations) and with greater mean velocity, particularly so during early stance (double-support period) when the *hy*A-F would have greatest influence.

METHODS

Participants

Twenty physically active, unilateral trans-tibial amputees (mean (SD) age 47.4 (12.5) years, mass 87.3 (13.5) kg, height 1.79 (0.06) m) took part, each giving written informed consent prior to their involvement. All had undergone amputation at least two years prior to participation (mean 11.85 (11.83) years, range 2 – 45 years) and all had used their current foot for at least six months. All participants habitually used a prosthetic foot with a fixed or elastically controlled articulating attachment (*hab*F). Twelve participants habitually used an Esprit foot (Esprit™, Chas. A. Blatchford and Sons Ltd., Basingstoke, UK). This foot is identical in design to the *hy*A-F, except that it uses a fixed attachment (figure 1). Of the other eight participants, five used a Multiflex, one a Flex-freedom, one an Elite and one a Seattle Litefoot. The study was conducted in accordance with the tenets of the Declaration of Helsinki and local bioethics committee approval was obtained.

Protocol and prosthetic intervention

Participants completed two blocks of 10 walking trials; one block was undertaken using their *hab*F and the other using a *hy*A-F. Block order was counter-balanced across participants and both blocks were conducted on the same day. Prior to completing the block using the *hy*A-F each participant's habitual prosthesis was altered by exchanging the existing foot for a *hy*A-F. All alterations were made by an experienced prosthetist, who was careful to ensure the two types of feet used had as close to the same alignment as possible. ~~To this end, everything about the prosthesis was kept constant (or near to constant as possible) when one foot type was exchanged for the other.~~ That is the socket, ~~and~~ suspension and alignment of the shank pylon were unchanged across foot types and each type of foot was attached to the distal end of the shank pylon with as close to the same alignment and set-up as possible. Thus before exchanging one foot for another, foot orientation and alignment of the attachment at the shank were noted and wherever possible maintained between foot conditions. When swapping from an Esprit (*hab*F) to an Echelon (*hy*A-F), or vice versa, the foot would naturally fall into the existing location and only shank length was adjusted (achieved by either shortening the shank pylon or replacing it with a longer one). When swapping one of the other types of *hab*F for a *hy*A-F, each foot's ideal alignment was used as the guiding criteria. Functioning (i.e. roll over characteristics) of each foot is optimal at its own ideal alignment, and using such alignment is therefore the fairest way to make comparisons between feet. Ideal alignment instructions were readily available from the respective manufacturers, and the experienced prosthetist making the adjustments was familiar with these instructions.

Once the *hy*A-F was fitted, participants walked both indoors and outdoors for a minimum of 45 minutes prior to data collection for accommodation. They negotiated ramps, slopes and stairs and walked over a variety of surfaces including pavements, grass verges and carpeted floors. During this period the settings which control the rates of articulation within the *hy*A-F (damping) were adjusted by the prosthetist until deemed to provide optimal function at self-selected, comfortable walking speed. The device has separate settings for plantar- and dorsi- flexion ranging from 1 [minimum] to 9 [maximum], equating to damping coefficients of 1.28 to 3.48 Nms/deg respectively. Participants completing trials using their *hab*F in the first block (block 1) completed these on arrival at the laboratory. For those completing trials using their *hab*F in the second block (block 2), the foot was refitted to their prosthesis following completion of block 1 (undertaken using the *hy*A-F), and the original length, set-up, and alignment of the prosthesis was restored. Participants were again given a familiarisation period, similar to that described above, in order to reacquaint themselves with their habitual prosthesis prior to data collection.

Data acquisition and processing

Participants walked in a straight line along a flat and level 8 m walkway at their freely-selected comfortable walking speed. Kinematic and kinetic data were recorded at 100 Hz and 400 Hz respectively using an eight camera motion capture system (Vicon MX, Oxford, UK) and two floor-mounted force platforms (AMTI, MA, USA) mounted within the floor of the walkway. A successful trial occurred when a 'clean' contact by the prosthetic foot was made with either of the two force platforms without any observable targeting or changes in stride pattern. During data collection, participants wore their own flat-soled shoes and 'lycra' shorts. Spherical, retro-reflective markers (all 14 mm diameter except markers placed onto the feet which were 9 mm diameter) were placed bilaterally on the following body landmarks (or equivalent locations on the prosthesis): acromion process, iliac crest directly above the greater trochanter, greater trochanter, medial and lateral femoral condyles, medial and lateral malleoli, posterior calcaneous, superior aspects of first and fifth metatarsal heads, distal end of second toe and pragmatically on the medial and lateral aspects of the mid-foot. Markers were also placed on the sternal notch, xiphoid process, and vertebrae C7 and T8. A head band was used to mount 4 head markers, and plate-mounted 4-marker clusters were worn on the thighs and shanks, whilst a skin-mounted 4-marker cluster was attached about the sacrum. Following 'subject'-calibration the markers on the acromions, knees and ankles were removed.

Labelling and gap filling of marker trajectories were undertaken within Workstation software (Vicon, Oxford, UK). The C3D files were then exported to Visual 3D motion analysis software (Version 4, C-Motion, Germantown, MD, USA), where a nine segment 6DoF model of each participant (Vanrenterghem et *al*., 2010) was constructed. Functional joint centres were created (as per Schwartz & Rosumalski, 2005) for both hips and knees and for the intact ankle. For the prosthetic limb a virtual 'ankle' centre was defined on the mid-line of the prosthetic shank at the same height as the contralateral intact ankle. This ensured a more consistent ankle definition between prostheses for valid shank rotation comparisons. These virtual landmarks were used to define the ends of the respective segments. Kinematic and kinetic data were filtered using a fourth order, zero-lag Butterworth filter with a 6 Hz and 20 Hz cut-off respectively. Initial contact (IC) and toe off (TO) were defined as the instants the vertical component of the ground reaction force first went above or below 20 N respectively. Double support (transfer onto prosthetic limb) was defined as being from prosthetic limb IC to contralateral limb TO. Single support was from contralateral TO to contralateral IC. When there were no kinetic data for the intact limb, IC and TO on the intact limb (which were used to determine single and double support for the prosthetic limb) were defined using kinematic data: IC was defined as the instant of prosthetic limb peak hip extension (De Asha et *al*., 2012) and TO as instant of peak posterior displacement of the intact toe marker relative to the pelvis (Zeni Jr. et *al*., 2008).

The global CoP and antero-posterior (A-P) COM displacements and velocities, and sagittal plane angular velocity of the prosthetic shank were exported in ASCII format for further analysis.

Data analysis

The following parameters were determined within Microsoft Excel (Microsoft, New York, NY, USA): average walking speed; mean and peak positive and peak negative (or minimum: if values remained positive) A-P CoP velocity; negative A-P CoP displacement; variability in CoP velocity across single support (indicating smoothness of forwards rotation); mean sagittal plane angular velocity of the prosthetic shank during the double support and single support phases. The first 5 ms of CoP data following IC were disregarded to avoid results being affected by any ‘foot-scuff’ during IC. To determine CoP negative displacement, the displacements occurring between each frame were first calculated and any negative displacements were then summed to give the total distance travelled by the CoP in the opposite direction to the direction of travel. Negative displacements in the CoP tended to occur during distinct periods (i.e. over several consecutive frames). CoP velocity variability was determined as the standard deviation in CoP velocity across single support. Angular velocity of the prosthetic shank was defined as the rate of rotation of the shank segment in the sagittal plane within the global co-ordinate system. Walking speed was defined as the mean forwards velocity of the COM during steady state walking (through the collection volume; length approximately 3 m). These parameters were calculated for each individual trial and then averaged across trials to give a mean value for each foot condition per participant.

Statistical analyses

Comparisons between foot conditions (*hab*F, *hy*A-F) were undertaken using 1-tailed, paired t-tests and by determining effect size differences. Effect size was calculated as Cohen's 'd' (Cohen, 1977). Statistical analyses were made using SPSS (Version 16, SPSS Inc, Chicago, IL, USA). The alpha level was set at 0.05. To check that any effects found between foot conditions were not specific to the type of *hab*F used by participants, we undertook a retrospective analysis whereby the participants were sub-divided into habitual Esprit users and non-Esprit users. Data were then analysed using mixed-design ANOVAs with sub-group (Esprit, non-Esprit) as between factor and foot condition (*hab*F, *hy*A-F) as within factor.

RESULTS

For all outcome variables there were no statistically significant differences between Esprit and non-Esprit users in terms of how they responded when switching to using a *hy*A-F (table 1). This confirmed the participants could be considered as one group, and thus the results described below are for the group as a whole.

The magnitude of the peak negative CoP velocity was significantly reduced from -0.153 (0.110) ms^{-1} when using the *hab*F to -0.043 (0.057) ms^{-1} when the *hy*A-F was used ($p < 0.001$, d = 0.9, table 1, also see figure 2a). The distance travelled posteriorly by the CoP reduced significantly from -0.022 (0.018) m using the *hab*F to -0.010 (0.008) m when using a *hy*A-F ($p = 0.001$, d = 0.6, table 1, also see figure 2b).

There were no significant differences in mean ($p = 0.24$) or peak ($p = 0.28$) anterior CoP velocity between foot conditions (figure 2a). CoP velocity variability across single-support was reduced from 0.273 (0.070) ms^{-1} when using the *hab*F to 0.201 (0.063) ms^{-1} when using the *hy*A-F ($p < 0.001$, d = 1.0, table 1).

Mean angular velocity of the prosthetic shank during the double support phase increased significantly from 94.5 (20.2) $°s^{-1}$ when using the *habF* to 101.7 (19.2) $°s^{-1}$ when using the *hy*A-F ($p < 0.001$, d = 0.3, table 1, also see figure 3). There were no significant differences in shank angular velocity between foot conditions during single support ($p = 0.37$).

Mean walking speed increased significantly from 1.12 (0.14) ms^{-1} (range 0.83 – 1.44 ms^{-1}) when using the *hab*F to 1.17 (0.15) ms^{-1} (range 0.84 – 1.44 ms^{-1}) using the *hy*A-F ($p = 0.001$, d = 0.3, table 1).

DISCUSSION

When walking with a non-articulated prosthetic foot the fore-foot is lowered to the floor following initial contact via a combination of heel deformation (creating simulated plantarflexion) and forward limb rotation (caused by bodyweight translating over the foot). When using a *hy*A-F, the lowering of the foot is also facilitated by the passive mechanical plantarflexion at the hydraulic device. The key findings of the present study were that use of a *hy*A-F significantly reduced or eliminated CoP posterior displacement, reduced the peak negative CoP velocity, reduced CoP velocity variability across single support, increased the mean forward shank rotational velocity during weight transfer onto the prosthesis, and increased overall walking speed. These findings support our hypotheses and indicate that use of the device led to bodyweight being transferred onto the prosthetic limb in a smoother, less faltering manner which allowed the COM to translate more quickly over the foot. This is likely why freely chosen walking speed was found to increase. In addition, participants reported the perception of having to ‘climb over’ their prosthesis was no longer present, which presumably was a consequence of the above findings.

Schmid et *al*. (2005) and Ranu (1988) have previously reported that ‘stalling’ of the CoP tended to occur under the hind-foot during early or mid-stance in trans-femoral and trans-tibial amputees respectively. In the present study the exact locations and timings of the disruptions to the anterior progression of the CoP were not consistent between participants with most participants displaying disruptions in CoP progression beneath the mid-foot in addition to the hind-foot. However, the location and timing of any CoP disruptions tended to be consistent within participants across foot conditions. This suggests that when and where CoP disruptions occur is not solely a function of the prosthetic foot used; rather it is an individual’s response to the foot and/or their style of walking. Low variability across the 10 repeated trials (see figure 2a) suggests such responses were consistent for each participant. In able-bodied gait the CoP A-P velocity pattern can be associated with the notion of the three ‘rockers’ of gait; with relatively high positive velocities during the first and third rockers (during which the foot rotates about the heel and toe regions respectively) and a slower, near constant velocity during the second rocker (during which the foot is relatively plantigrade and the ankle becomes the rocker, see figure 2a). In general both the *hab*F and *hy*A-F devices were able to mimic, albeit to differing degrees, the first two rockers with regard to CoP velocity. However the first rocker (reflected by a short duration rapid increase in CoP velocity, figure 2a), was temporally delayed compared to that in able-bodied gait, particularly so when using the *hab*F. This delay was likely a consequence of the compression/deformation of the prosthetic heel keel needed to allow the foot to become plantigrade. Such ‘early stance’ CoP disruption corroborates previous findings that have indicated the CoP becomes ‘stalled’ under the prosthetic hindfoot (Schmid et *al*. 2005). This delay was reduced, but not removed when using the *hy*A-F. There were also fewer CoP velocity fluctuations during the second rocker period (single support) when using the *hy*A-F compared to *hab*F (as evidenced by the reduced variability in CoP velocity), supporting the hypothesis that CoP trajectory disruption would be reduced. As single-support represents the period when there are no propulsive or braking forces applied by the contralateral (intact) limb, fewer fluctuations in CoP velocity during this period reflect a more uniform transfer of bodyweight over the prosthesis. Finally, it is apparent that participants were unable to facilitate generating a short duration rapid increase in CoP velocity during the third rocker as seen in able-bodied controls irrespective of which prosthetic foot was being used. This is due to the lack of active plantarflexion via concentric muscle action prior to TO and highlights the major limitation of all passive prosthetic feet.

Due to the counter-balanced experimental design and because of the methodological limitations associated with speed-controlled studies and the difficulty in generalising findings from such studies to the natural environment (Wilson, 2012) we decided not to control walking speed. Instead participants were asked to walk at a speed they perceived to be customary. As highlighted above, this freely chosen walking speed was found to be significantly greater when participants used the *hy*A-F. Consequently, in order to establish whether the CoP trajectory changes found when participants used a *hy*A-F were simply due to an increase in walking speed rather than the functioning of the device, we retrospectively investigated the relationship between trial walking speed and CoP progression. This analysis highlighted that there was no significant relationship between walking speed and the magnitude of peak negative CoP displacement, mean CoP velocity or peak positive CoP velocity irrespective of foot type ($R^2 \leq 0.015$, $p \geq 0.1$). However, peak negative CoP velocity was significantly related to walking speed when using the *hab*F (r = -0.1672, $R^2 = 0.0280$, $p = 0.025$), but not when using the *hy*A-F ($p = 0.35$). This indicates that when using the *hab*F the velocity of the negatively directed CoP excursion was greater (i.e. increased in negative direction) in trials completed at higher walking speeds. We can only speculate about the cause of this relationship. It is possible that at higher walking speeds the *hab*F had a tendency to 'bottom out' during loading of the heel-keel i.e. period of weight acceptance, and this then affected (delayed) how weight was transferred onto the fore-foot keel as the COM progressed forwards over the foot. Given that walking speed was found to be unrelated to any of the CoP measures when using the *hy*A-F, this suggests the findings, indicating use of the *hy*A-F attenuated CoP trajectory fluctuations, resulted from the functioning of the foot rather than simply being due to an increase in customary walking speed when using this foot.

Irrespective of foot type, the mean forwards angular velocity of the prosthetic shank was significantly higher during double support (weight transfer onto the prosthetic limb) compared to single support, and velocities during double support became significantly increased when using the *hy*A-F, supporting the hypothesis that shank angular velocities would be increased during early stance. This may have contributed to the significant increase in overall walking speed when using the *hy*A-F. A systematic review of the variables used in amputee gait research highlighted that self-selected comfortable walking speed was the most often reported parameter (Sagawa Jr. et *al*., 2011), which reflects the importance of walking speed as a measure of overall gait quality. Increasing walking speed has been found to decrease temporal asymmetries in amputee gait (Nolan et *al*., 2003) and is positively correlated with amputees' self-perception of gait quality (Miller et *al*., 2001). A previous study that compared use of ESR and non ESR feet with and without an elastic ankle articulation device (Zmitrewicz et *al*., 2006) found no difference in walking speed across foot and ankle device conditions. This suggests that ankle articulation alone does not result in an increase in walking speed. It has recently been demonstrated that use of the same type of *hy*A-F device as used in the present study led to a reduction in in-socket pressure in trans-tibial amputees during ambulation at self-selected speeds (Portnoy et *al*., 2012). This suggests that comfort might be increased when using the device and it may be this increased comfort which also facilitates higher walking speeds.

There were no statistically significant differences between participants who habitually used an Esprit foot and those habitually using non-Esprit feet across the foot conditions (*hab*F, *hy*A-F). This indicates that the effects observed as a result of using the *hy*A-F can be generalised across all participants irrespective of which type of habitual foot they used. However, while there were no significant differences between habitual Esprit and habitual non-Esprit users, effect size differences between foot conditions (*hab*F, *hy*A-F) tended to be larger for the Esprit users than for the non-Esprit users (e.g. reduction in negative CoP displacement; Esprit, d = 0.8, non-Esprit, 0.5). This may either reflect limitations in the functional performance of the Esprit foot or improved functionality for the non-Esprit feet. Five of the eight non-Esprit users used a Multiflex (Chas. A. Blatchford and Sons Ltd., Basingstoke, UK) device which allows elastically controlled articulation at the ‘ankle’ attachment. Indeed, the Multiflex foot-ankle provides up to 15° of sagittal plane articulation which is more than the *hy*A-F does. However, as highlighted above, differences between foot conditions (*hab*F, *hy*A-F) were significant for both the non-Esprit (predominantly Multiflex) and Esprit users. This suggests that the observed effects of using a *hy*A-F were due to the hydraulically dampened nature of the articulation rather than solely the magnitude of the articulation provided. The *hy*A-F provides passive dampened (time-dependent) resistance which slows and thus temporally extends the period during which articulation occurs compared to elastically controlled devices. With elastically controlled devices, such as a Multiflex, articulation is permitted via deformation at the point of attachment (e.g. by use of a rubber snubber). The rate of articulation is governed by the stiffness of the snubber and is, by and large, time-independent. These devices will reach their limit of articulation very quickly when loaded at which point they will act more like a rigid device. This would explain why there were no significant differences between Esprit users and non-Esprit (predominantly Multiflex) users.

One potential limitation of this study is that participants had minimal experience of the *hy*A-F prior to data collection sessions, and although the data collection blocks for each foot condition were completed in a counter-balanced manner to minimise order effects, we cannot say with certainty that familiarisation had no effect on results. However, the findings of the present study suggest mechanical benefits of using a *hy*A-F compared to either a non-articulating or elastically controlled articulating foot-ankle device, and it is difficult to see why familiarisation would negate such benefits. Another limitation is that all participants in the study were active unilateral trans-tibial amputees, and future work should thus investigate whether use of a *hy*A-F would have similar effects in trans-femoral amputees and/or in those who are less active. The *hy*A-F weighs approximately 400 g more than the same type of foot (i.e. Esprit) without the hydraulic device so there may be an extra metabolic cost involved with its use. This was not measured as part of the present study so should be investigated as part of any future study.

CONCLUSION

Use of the hydraulic ankle-foot device reduced or eliminated the backwards directed CoP displacement, reduced CoP velocity fluctuations beneath the prosthetic foot and allowed the prosthetic shank to rotate over the foot quicker during double support. These changes were associated with an increase in self-selected comfortable walking speed. This suggests that such a device may be functionally beneficial for active amputees. In addition, this study has highlighted that among other measures, which aim to quantify comparative performance of prosthetic feet, the examination of the CoP progression beneath the prosthetic foot is a useful tool.

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Conflict of interests – None.

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Table(s)

Table 1. Group mean (SD) CoP trajectory measures, shank velocities and walking speeds when walking with *hab*F and *hy*A-F. Participants in Sub-G1 habitually used an Esprit foot and participants in Sub-G2 habitually used a range of other types of feet (see text for detail). Measures that differed significantly when switching to an *hy*A-F are shown in bold. Where differences are significant effect sizes (d) are provided (in *italics*).

		negative CoP displaceme nt (m)	maximum negative CoP velocity (ms^{-1})	maximum positive CoP velocity (ms^{-1})	mean CoP velocity (ms^{-1})	mean CoP velocity variability (Single Support) (ms^{-1})	shank mean angular velocity (Double Support) ($°s^{-1}$)	shank mean angular velocity (Single Support) ($°s^{-1}$)	Walking speed (ms^{-1})
*hab*F	ALL	-0.022 (0.018)	-0.153 (0.110)	2.392 (0.892)	0.365 (0.041)	0.273 (0.070)	94.5 (20.2)	66.5 (9.9)	1.12 (0.14)
	SubG1	-0.026 (0.019)	-0.210 (0.092)	2.607 (1.043)	0.361 (0.036)	0.283 (0.060)	91.8 (23.2)	66.6 (10.4)	1.11 (0.15)
	SubG2	-0.016 (0.013)	-0.066 (0.073)	2.072 (0.504)	0.371 (0.051)	0.267 (0.080)	98.7 (15.3)	66.5 (10.0)	1.14 (0.14)
*hy*A-F	ALL	**-0.010 (0.008)** ***0.6***	**-0.043 (0.057)** ***0.9***	2.305 (0.890)	0.370 (0.043)	**0.210 (0.063)** ***1.0***	**101.7 (19.2)** ***0.3***	66.2 (8.8)	**1.17 (0.15)** ***0.3***
	SubG1	**-0.010 (0.008)** ***0.8***	**-0.062 (0.066)** ***1.4***	2.535 (1.006)	0.378 (0.036)	**0.212 (0.073)** ***1.5***	**100.6 (21.6)** ***0.3***	66.7 (9.3)	**1.17 (0.15)** ***0.3***
	SubG2	**-0.009 (0.008)** ***0.5***	**-0.014 (0.021)** ***0.7***	1.960 (0.577)	0.358 (0.055)	**0.204 (0.050)** ***0.8***	**103.3 (16.0)** ***0.2***	65.3 (8.4)	**1.18 (0.14)** ***0.2***

Figure 1
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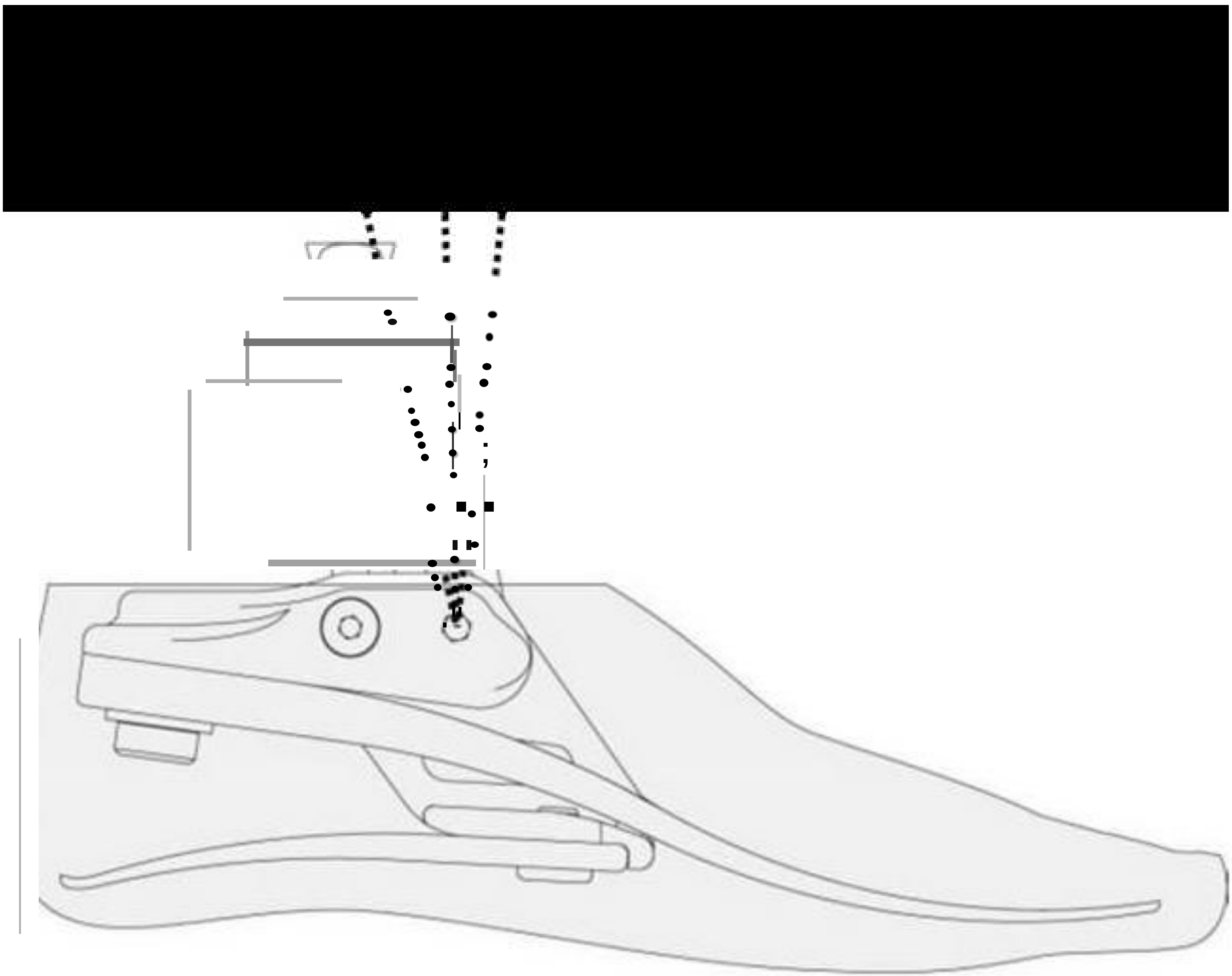

Figure2a
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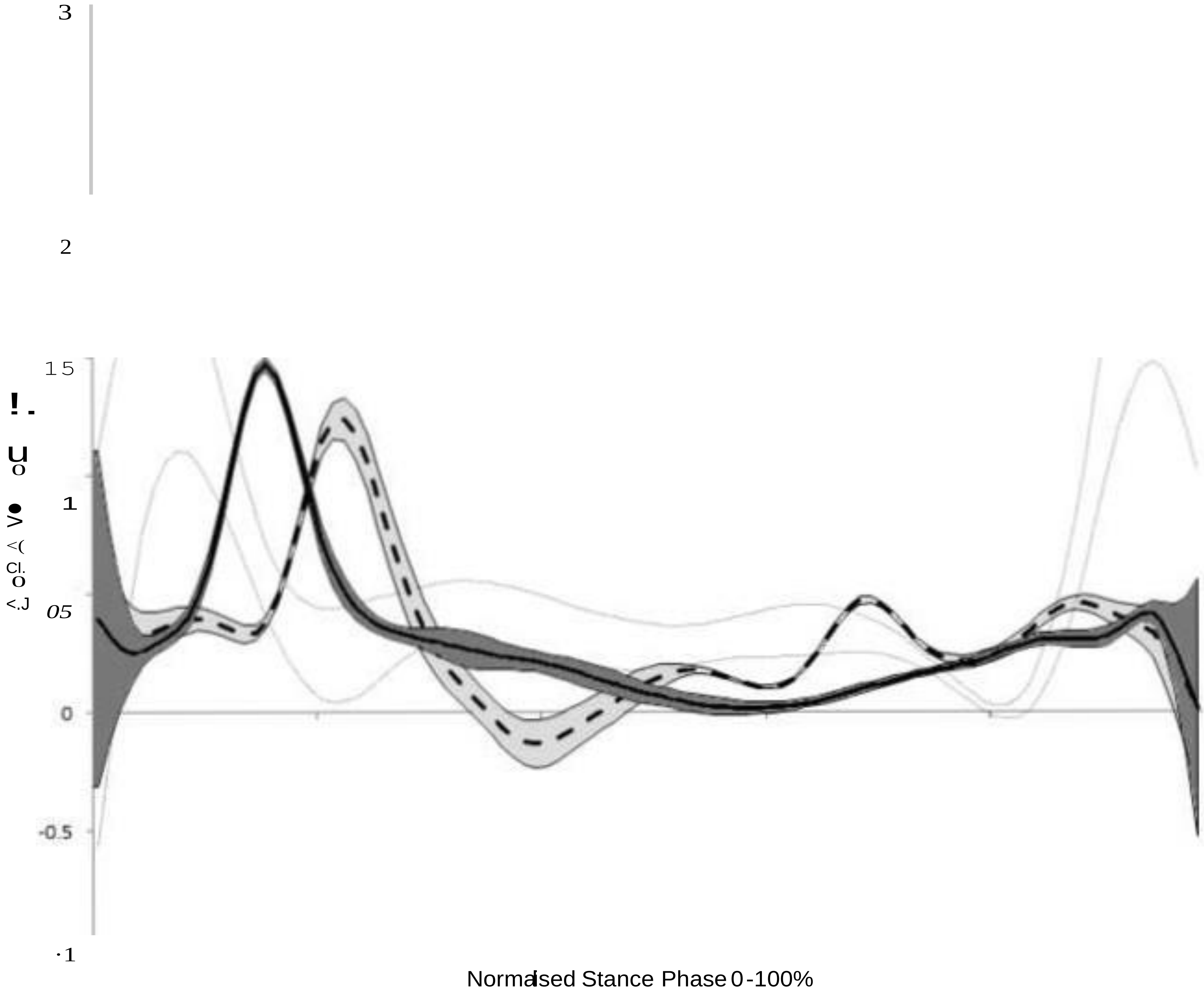

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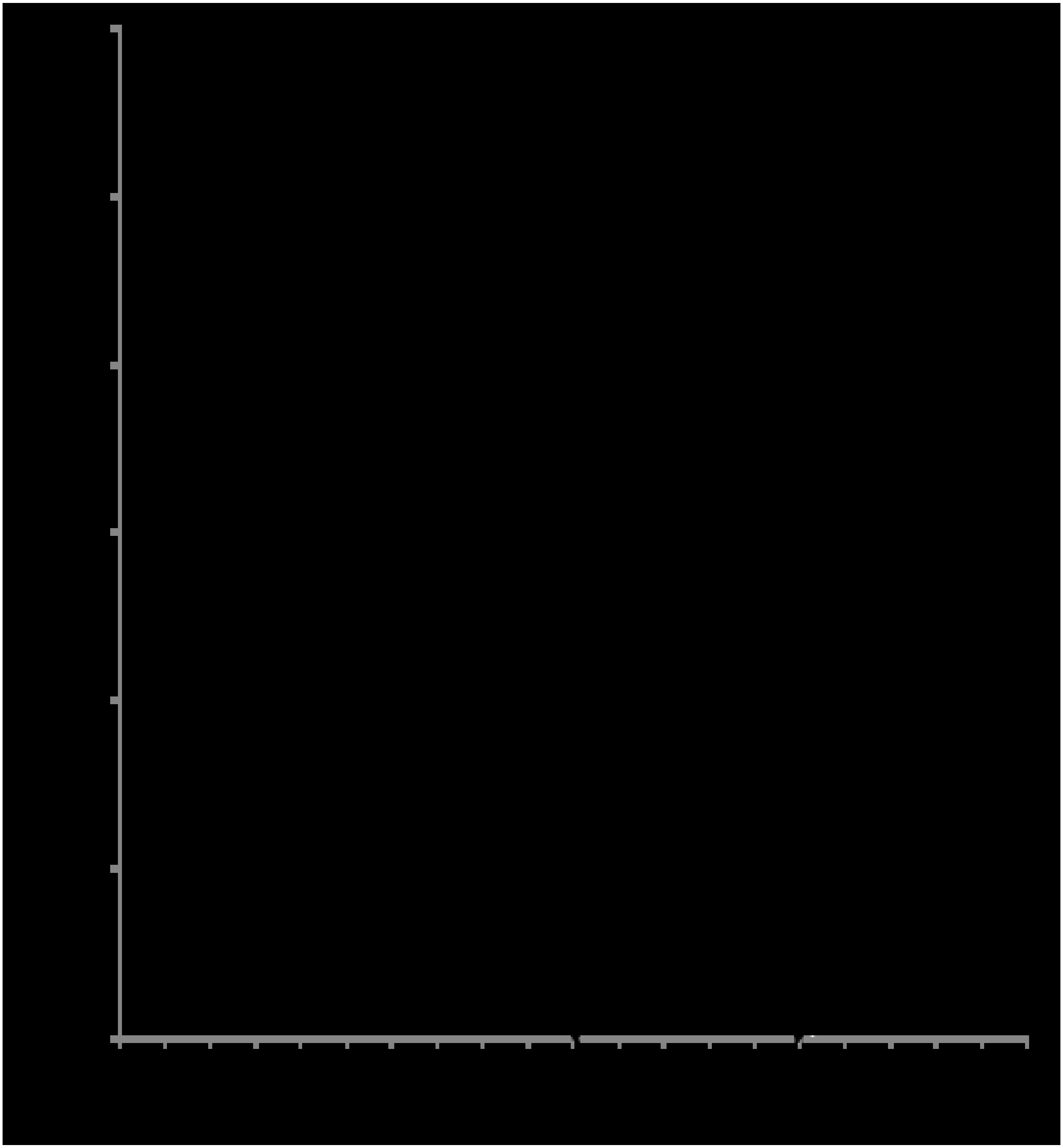

Figure3
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