THE EFFECT OF WALKING SPEED, SLOPES, AND STAIRS ON DYNAMIC MEAN ANKLE MOMENT ARM

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Introduction

Understanding the control mechanism of the human foot-ankle complex can inform the design of novel foot prostheses. To improve the adaptation of these designs to different activities such as ramp and stair descent, it is important to observe the way ankle control changes in a non-impaired human limb. In this study, we observe the changes in dynamic mean ankle moment arm (DMAMA) [1] when performing different activities. DMAMA describes the way the ankle controls the location of force interaction with the ground (moment arm). This summative measure accounts for temporal, spatial, and directional variations in the applied force in a single value. This makes DMAMA a useful target measure for semi-active prostheses that adjust only once per stride.

Methods

Eight unimpaired adults (age: 30±6 years; mass: 65±13 kg; foot length: 28±2 cm; mean ± SD) provided written informed consent to participate in this study. Participants performed at least 8 trials of each of the 7 activities. Walking speeds were self-selected. Kinematics were collected using a 12-camera motion capture system and ground reaction forces using multiple force plates: 6 embedded in a level walkway, 2 in a 5° incline ramp, and a force plate stairway with 3 steps, the last step forming a part of the top platform.

We used Visual3D (C-Motion, Inc.) to filter the force and kinematic data, determine foot contact with force plates, and calculate ankle moment and ground reaction force/moment. We used a custom MATLAB script to determine heel contact (HC) and toe-off (TO) and calculate DMAMA during stance phase according to Equation 1 where M is the ankle moment and F is the sagittal ground reaction force [1]. DMAMA values were subsequently normalized to participant foot length.

$$DMAMA = \frac{\int_{HC}^{TO} M\,dt}{\left\| \int_{HC}^{TO} F\,dt \right\|} = \frac{\bar{M}}{\bar{F}} \qquad (1)$$

To assess the difference in DMAMA across different activities, we fit a linear mixed model (LMM) to the means of each subject's DMAMA for each category: speed, slope, and stairs. We included DMAMA as the dependent variable, the activities as the fixed effects and the participant as a random effect. We define sensitivity in table 1 as the regression coefficient from this LMM and each change in activity corresponds to an increment of one (e.g., for speed: slow (-1), normal (0), fast (1)). Critical α was set to $p<0.05$.

Results and Discussion

As walking speed increased, DMAMA decreased as shown in figure 1 and table 1. This result matches previous work by Adamczyk [1], who also found that DMAMA tended to move towards the heel (decrease) as walking speed increased. Higher DMAMA at slower walking speeds may be due to an increased plantar flexor moment attempting to slow the body's progression [2]. We observed an even sharper decrease in DMAMA when participants ascended stairs versus descended them.

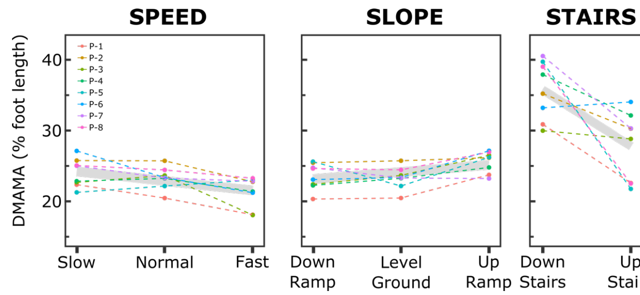

Figure 1: Trends in DMAMA across activities. The thick grey lines show the subject-independent data fit as a result of the linear mixed model. Colored markers connected by dashed lines represent each participant's average in the different categories.

Table 1. Subject-independent regression – LMM

Fixed Effect	*Sensitivity (DMAMA per increment)*	*p-value*	R^2
Walking Speed	-1.34	0.002	0.49
Slope	+1.01	0.010	0.60
Stairs	-8.00	0.003	0.46

Table 2. Mean DMAMA (% foot length) by Activity

Activity	*Mean± SD*	*Activity*	*Mean± SD*
LG Walking:		Down Ramp (-5°)	23.6 ± 2.60
Slow	23.9 ± 2.85	Up Ramp (5°)	25.5 ± 2.20
Normal	23.3 ± 2.28	Up Stairs	27.8 ± 5.48
Fast	21.1 ± 2.54	Down Stairs	35.9 ± 4.99

As the ground incline changed from negative to positive, we observed a significant shift forward in DMAMA. This is consistent with prior work by Leestma et al. [3] who showed a similar increase in DMAMA with increasing ground slope in participants with amputation using an experimental foot-ankle prosthesis. While the sensitivities in table 1 may appear small, DMAMA moving 1-8% foot length per increment, it is important to note that typical DMAMA values fall within a relatively narrow numerical range [1].

Significance

The results of this study can be used to design more biomimetic assistive devices such as prostheses, exoskeletons and orthoses. Using approaches such as varying prosthetic keel stiffness to tune DMAMA, semi-active prostheses can alter the biomechanics of a person's gait [3]. Target DMAMA values, such as the ones presented in this work, could be used to inform the control algorithms of novel prostheses, allowing them to adapt to different activities. DMAMA could also be used as a performance metric in fully robotic devices to evaluate and adjust their continuous ankle torque controllers. Having the ability to adapt to different speeds and terrains may improve the experience of prostheses users, e.g., by reducing maladaptive socket torques.

Acknowledgments

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References

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