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Metatarsophalangeal Joint Dynamic Stiffness During Toe Rocker Changes With Walking Speed

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Dynamic joint stiffness (or simply "stiffness") is a customization criteria used to tune mechanical properties of orthotic and prosthetic devices. This study examines metatarsophalangeal (MTP) joint stiffness during the toe-rocker phase of barefoot walking and establishes baseline characteristics of MTP joint stiffness. Ten healthy individuals walked at 4 speeds (0.4, 0.6, 0.8, and 1.0 statures·s−1) over level ground. MTP sagittal plane joint angles and moments were calculated during the toe-rocker phase of stance. Least-squares linear regressions were conducted on the MTP moment versus angle curve to determine joint stiffness during early toe rocker and late toe rocker. Multilevel linear models were used to test for statistically significant differences between conditions. Early toe rocker stiffness was positive, while late toe rocker was negative. Both early toe rocker and late toe rocker stiffness increased in magnitude significantly with speed. This study establishes baseline characteristics of MTP joint stiffness in healthy walking, which previously had not been examined through a range of controlled walking speeds. This information can be used in the future as design criteria for orthotic and prosthetic ankle and ankle–foot devices that can imitate, support, and facilitate natural human foot motion during walking better than existing devices.

Keywords: dynamic joint stiffness, gait, foot and ankle

Dynamic joint stiffness (hereafter referred to simply as "stiffness") is a descriptive metric commonly used for numerous joints in the body, particularly lower-limb joints.[1–3] It is most often defined as the slope of the linear regression of a particular joint's net moment versus angle throughout all or part of a defined motion such as walking or running.[1] This measure provides a surrogate for the stiffness of a rotational spring about that joint[1,4,5] and has been used to drive the design of biomechanical assistive devices. Ankle stiffness has been an important parameter for customizing the bending stiffness of passive ankle–foot orthoses and prostheses.[6–8] Recent work in running biomechanics has highlighted the importance of foot stiffness in the context of energy storing-and-releasing footwear.[9–11] In prosthetics, forefoot stiffness has been shown to affect contra- and ipsilateral lower-limb dynamics and play an important role in user comfort.[12–15] Finally, it has been shown that elastic foot orthoses can provide added support to the natural foot structures (plantar aponeurosis, intrinsic muscles) that store and release energy about the metatarsophalangeal (MTP) joints[16,17] and can possibly store and release energy that would otherwise be dissipated during stance.[18,19]

Previous work examining MTP joint dynamics includes methodological studies (marker set development, joint definitions) of the joints intrinsic to the foot[20–23] and those joints' relation to intrinsic foot structures like the plantar aponeurosis and intrinsic foot muscles.[16,24] There is a wealth of research regarding MTP stiffness, longitudinal foot stiffness, and shoes' effects on those metrics during running,[25–28] which indicate that MTP and foot stiffness increase with running speed. However, a gap in knowledge exists regarding MTP stiffness during walking. Farris et al[24] studied the effects of nerve blocks on intrinsic foot muscle activation and MTP stiffness during walking at a single speed, and Sanchis-Sales et al[23] studied MTP stiffness during walking, but only during stance phases that were defined by ankle joint mechanics, not MTP joint mechanics. These 2 studies, however, lie within a broader gap in the literature regarding MTP stiffness. There have been no attempts to the authors' knowledge to examine the effect of walking speed on MTP stiffness regions defined by MTP joint mechanics.

While the running field's interest in this topic has primarily been due to the potential for elastic foot plates to improve shoe performance, our interest comes from a rehabilitation perspective. Certain patient populations with foot or ankle injuries or impairments might benefit from orthotic or prosthetic devices with flexible foot plates that can substitute for lost lower extremity function. This notion is based on evidence from individuals who have had a stroke that quantitatively prescribing ankle–foot orthosis stiffness such that it, when added to the user's decreased stiffness-generating capability, sums to the stiffness of a healthy, age-matched, speed-matched individual.[29] In that case, it is first necessary to know what normal, healthy ankle stiffness is and how much difference there is between a patient and that healthy baseline. Quantifying typical MTP stiffness, and how it changes across a range of gait activities, could provide a target stiffness goal when optimizing devices with tuned forefoot stiffness for patients. If individuals have a foot or lower-limb impairment that prevents them from generating sufficient MTP stiffness, and we know what stiffness they *should* be generating, a flexible foot plate could be tuned to match that stiffness discrepancy.

The purpose of this study was to provide a baseline for MTP stiffness during terminal stance and understand how MTP stiffness changes with walking speed in young, healthy individuals. Knowing baseline healthy MTP stiffness will be crucial in designing and optimizing assistive devices that can either support an existing foot (orthosis) or behave like a natural foot (prosthesis) in a way that is beneficial to the user across multiple speeds. We hypothesized that

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MTP joint stiffness would increase in magnitude with speed during terminal stance.

Methods

Participants

Ten young, healthy individuals (average [SD]: age = 24.2 [2.86] y, height = 1.71 [0.07] m, mass = 69.63 [10.71] kg, 6 women/4 men). Exclusion criteria included: any history of chronic neuromusculoskeletal disease (eg, stroke, cerebral palsy), any minor lower-limb musculoskeletal injury within the last 6 months (eg, grade 1 sprains, plantar fasciitis), or any history at any time of injuries that included fully or partially torn tendons or ligaments (eg, grade 2–3 sprains, ACL tears, Achilles tendon tears). Subjects were recruited from the University of Delaware during 2011. The time between recruitment and data collection for any subject was less than 2 weeks. Experimental protocols were approved by the University of Delaware's Institutional Review Board, which ensures that human subject research at the university abides by the Declaration of Helsinki and the Nuremberg Code. All subjects provided written informed consent prior to participating.

Experimental Protocol

Subject anthropometrics were recorded for biomechanical model building. Static calibration trials were recorded and then subjects traversed an approximately 20-m-long level walkway while wearing a 6-degree-of-freedom marker set on their pelvis and lower limbs.[30] Most relevant to the analysis in this study were a cluster of 3 tracking markers placed on the dorsal foot, a cluster of 4 markers located on the lateral shank, and markers to identify joint centers at the lateral malleolus (ankle) and lateral femoral epicondyle (knee) (Figure 1). MTP joint centers were identified at the second metatarsal head, and 3D measurements from the most medial tracking marker on the foot to the MTP joint center were recorded for 3D reconstruction of the MTP joint center during analysis. This marker set has been shown to be reliable and valid for identifying joint center locations and net joint moments across a wide range of walking speeds, even with up to 10 mm variation in placement of knee joint center markers.[30] Ground reaction forces were measured at 720 Hz by 4 force plates (AMTI) embedded in the runway, flush with the floor surface. 3D kinematics was captured at 240 Hz with a 6-camera motion capture system (Motion Analysis Corp).

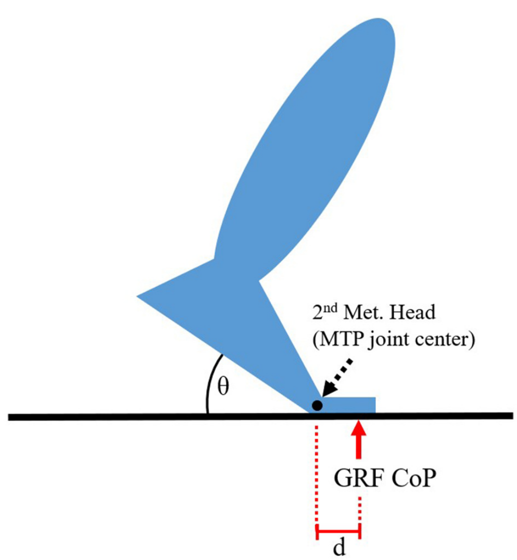

Figure 1 — Representation of the measurements used to calculate MTP joint stiffness during the forefoot rocker. The MTP joint angle is calculated as the foot-to-floor angle (θ). The moment arm d was calculated as the distance between the MTP joint center (black dot) and the GRF CoP, measured from the force plate. GRF CoP indicates ground reaction force center of pressure; MTP, metatarsophalangeal.

Subjects walked barefoot at 0.4, 0.6, 0.8, and 1.0 statures·s^{-1} with a tolerance of ±0.02 statures·s^{-1} (statures·s^{-1} = meters·s^{-1}·subject height in meters^{-1}). 0.8 statures·s^{-1} represents a comfortable walking speed for most healthy individuals,[31] so the range of speeds broadly encompasses very slow to moderately brisk walking. Speed was set in this manner because it allowed for a controlled, standard way for every subject to modulate their speed in the same way, rather than simply telling subjects to walk "slow, medium, and fast," which could result in excessive intrasubject and intersubject variability. On average, these speeds correspond to approximately 0.68, 1.03, 1.37, and 1.71 m·s^{-1}, respectively, for this study. The order of the speeds was randomized. To monitor speed, infrared gates were placed on either side of the force plates, 10 m apart. After each trial, a verbal cue was given to the subjects to slow down, speed up, or maintain their speed for the next trial. For every speed condition, 4 good trials per limb were collected. A good trial was defined as one in which the subject's speed was within the tolerance for that condition and only one of their feet was in complete contact with a single force plate for all of stance. Subjects did not explicitly practice the speed of each condition prior to data collection; however, it often took approximately 5 trials at a new condition for subjects to achieve the correct speed with clean force plate strikes. Neither limb was explicitly chosen to be collected first. Rather, the first foot with which a subject cleanly struck a force plate had that data collected first, which did not correlate with a subject's footedness.

Data Processing and Analysis

All data were postprocessed using Cortex (Motion Analysis Corp), and inverse dynamics were computed with Visual 3D (C-Motion Inc). Kinematic and kinetic data were filtered at 6 Hz and 25 Hz, respectively, using a fourth-order zero-lag Butterworth filter. All data were analyzed only during the toe-rocker phase of stance, since negligible MTP joint bending was observed during early and mid-stance. The toe rocker, presently defined, began when the anterior/posterior position of the center of pressure crossed the MTP joint center (Figure 2). The MTP joint center was defined as the location of the second metatarsal head (MTH), which is a commonly used method.[9,22] Our marker set did not have an explicit second MTH marker, so the second MTH's distance from one of the foot cluster markers was measured in all 3 dimensions instead. These measurements were used to recreate the second MTH location in our model. A comparison between this method and others, including using the first MTH, the fifth MTH, the midpoint between the first and fifth MTHs, and a "sliding" MTP joint center, showed that using the second MTH method resulted in MTP joint moments closest to the sliding method.[32] Using the first or fifth MTH as the MTP joint center definition greatly overestimates MTP joint moment values.[32]

To calculate MTP joint stiffness, the joint angle and net moment were first calculated. The MTP joint angle was defined as the overall foot-to-floor angle. In actuality, the MTP joint angle

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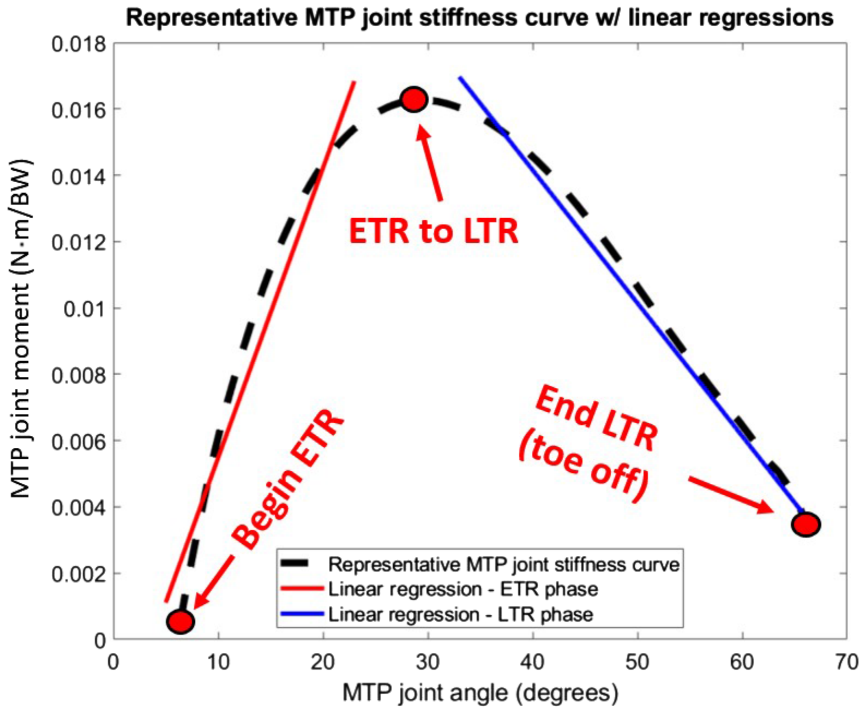

Figure 2 — Representative MTP joint stiffness curve (black dash) with linear regressions on ETR (solid line on left, positive slope) and LTR (solid line on right, negative slope) subphases. The forefoot rocker phase begins at the left-hand side of the plot and progresses to the right-hand side. MTP joint angle steadily increases, while MTP joint moment rises, peaks, and falls. BW indicates body weight; ETR, early toe rocker; LTR, late toe rocker; MTP, metatarsophalangeal.

would be the supplement of the foot-to-floor angle. Since the relevant data were how foot-to-floor angle *changed* in relation to the MTP moment (Figure 2), any offset in foot-to-floor angle versus MTP joint angle was not relevant. The net moment about the MTP joint was defined as the anterior/posterior distance between the MTP joint center and the center of pressure multiplied by the vertical GRF (normalized ground reaction force). GRF was normalized by subject body weight, and as such, MTP joint moments were normalized by subject body weight as well (not mass). The net moments about the MTP joint, GRFs, marker trajectories, and center of pressure data were all normalized in time to percent toe-rocker phase. MTP joint angles and MTP joint moments were then exported from Visual 3D, and further analysis was conducted with a custom MATLAB (MathWorks Inc) code.

The toe-rocker phase was divided into 2 subphases, which we will call early toe rocker (ETR) and late toe rocker (LTR). The ETR phase was defined as the period from the beginning of the toe-rocker phase until the peak net plantar flexion moment about the MTP joint (Figure 3). The LTR phase was defined as the period from peak net plantar flexion moment about the MTP joint until toe-off. For each foot strike and each toe-rocker subphase, a linear least-squares best-fit regression on net MTP joint moment versus MTP joint angle was computed, and the first-order coefficient (slope) was used to define the MTP joint stiffness. The first and last 5% of both phases were omitted from the regressions to better isolate the linear portions of MTP joint stiffness, following Sanchis-Sales et al.[23] The regressions took the form: $\tau = k\theta + constant$ where τ is the net MTP joint moment (N·m·body weight^{-1}) for each condition and phase of the toe rocker, and θ is the MTP joint angle and the constant term represents the applied moment at the neutral position, which corresponds to a foot and toes that are both flat on the ground. To clarify, MTP stiffness was calculated with a linear regression of MTP joint angle on net moment, which resulted in k being a bending or torsional stiffness about the MTP joint center in the sagittal plane. It is not a linear stiffness, and every mention of the word "linear" in this paper only refers to the linear *regression* used, not any type of *linear* stiffness. Put another way, $k = \Delta\tau/\Delta\theta$. Data were not averaged across subject, speed, or limb during analysis to ensure intersubject and intrasubject variability was not washed out.

Statistical Analysis

To evaluate our hypothesis, stiffness values were analyzed for differences between speed conditions for each toe-rocker subphase. Stiffness values were analyzed using 3 multilevel linear models (MLM) in SPSS software. Broadly, these models answer the question: "Does speed play a significant role in predicting MTP joint stiffness during the ETR and LTR portions of terminal stance, while also accounting for subject and dominant/nondominant limb?" Limb dominance was determined as the limb with which the subject would kick a ball. MLMs are a type of multiple regression analysis, so to avoid confusion, the term "regression" and "regression coefficients" used in this paper will always refer to the moment/angle regressions performed on each trial, not the MLMs.

MLMs include random and fixed effects. Broadly, they are the best analysis option for hierarchically nested data[33,34] (eg, dominant/nondominant leg within subject). By accounting for random and fixed effects, the data become homogenous.[35,36] As such, all trials within a given condition, regardless of subject or limb, can be directly compared with one another. This yields 80 homogenous data points. The dependent variables were the polynomial coefficients for each subphase of toe rocker. The fixed-effect predictors included a

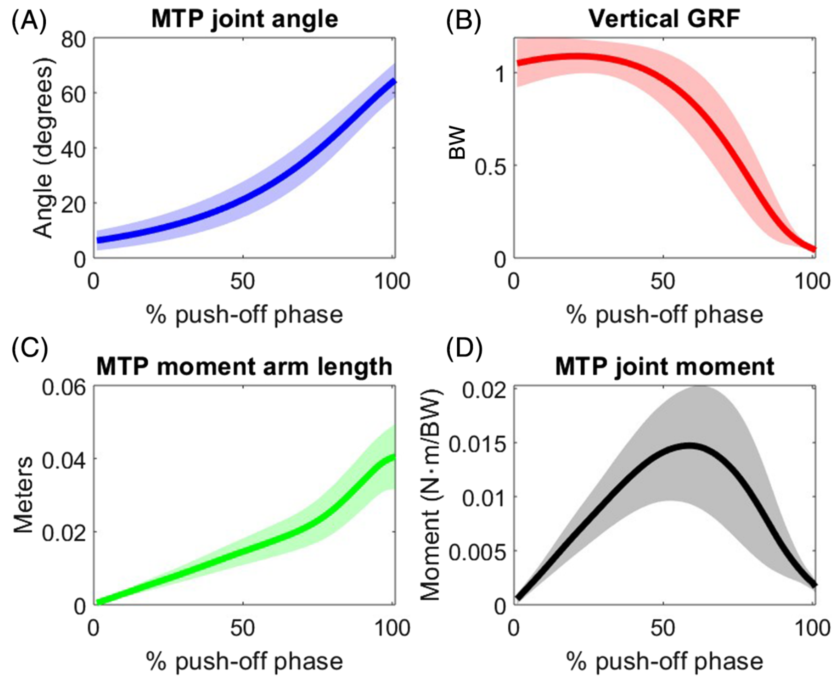

Figure 3 — Average (solid line) and ±1 SD (shaded area) across all subjects of (A) MTP joint angle (top left), (B) normalized GRF (normalized by BW; top right), (C) MTP joint moment arm length (bottom left), and (D) MTP joint moment (bottom right) during the push-off phase for the 0.8 statures·s^{-1} walking speed condition. Similar patterns and variations were seen for other walking speed conditions. BW indicates body weight; GRF, ground reaction force; MTP, metatarsophalangeal.

dummy-coded variable to control for dominant/nondominant limb, codes for each subject to control for intersubject variability, and 3 dummy-coded variables that compared speeds. Each model used 1 walking speed condition as the "base," against which all other conditions were compared. This way, every condition was compared directly to every other condition for each outcome measure in Table 1 (except r^2 and root mean square error of approximation). For example, the model which used the 0.4 statures·s^{-1} condition as its base compared the stiffness values of that condition with the 0.6, 0.8, and 1.0 statures·s^{-1} conditions directly, but did not compare the 0.6 statures·s^{-1} condition with the 0.8 statures·s^{-1} condition. Thus, 3 models with 0.4, 0.6, and 0.8 statures·s^{-1} as base conditions, respectively, will yield direct comparisons between every condition, plus redundancy. The MLMs used maximum-likelihood estimation with variance components as the variance–covariance error structure. Restricted maximum-likelihood estimation was not utilized because it may provide better estimates of *population* means but doing so makes assumptions about the variance about that mean, an assumption we were not comfortable with. Variance components were chosen as the covariance structure because that is an appropriate assumption for this repeated measures–formatted experimental design (same subjects performing the same task under different conditions). All P values were 2-tailed, and significance level was $\alpha = .05$.

Results

For the regressions, the average correlation of determination was high (mean [SD]) ($r^2 > .946$ [<.025]), and regression root mean squared error (RMSE) values were all low (<4.49e^{-6} [<8.58e^{-7}]) across all speeds (Table 1). These metrics indicate that MTP joint stiffness during ETR and LTR are sufficiently linear to be appropriately approximated by linear regression. Examples of each component contributing to the overall stiffness calculation can be seen in Figure 4. As expected, peak MTP joint angle, peak GRF, and peak MTP moment all increased significantly with walking speed.

Averaged across all trials, ETR stiffness steadily increased with walking speed from 7.0e^{-4} (3.7e^{-4}) to 1.09e^{-3} (3.8e^{-4}) N·m·deg^{-1}·body weight^{-1} at 0.4 statures·s^{-1} to 1.0 statures·s^{-1}, respectively (Table 1, Figure 4). The MLMs indicated these differences in ETR stiffness were significant between all conditions ($P < .014$) except 0.4 to 0.6 statures·s^{-1} ($P > .066$). LTR stiffness became gradually more negative as waking speed increased, from −2.56e^{-4} (1.3e^{-4}) to −5.10e^{-4} (1.9e^{-4})N·m·deg^{-1}·body weight^{-1} at 0.4 statures·s^{-1} and 1.0 statures·s^{-1}, respectively. The MLMs indicated the differences in LTR stiffness were significant between all conditions ($P < .003$). Additionally, dominant/nondominant limb was shown to be a significant predictor for both ETR and LTR stiffness ($P < .05$).

The constant term was nearly zero for ETR for all speeds and exhibited no significant differences between conditions ($P > .081$, Table 1). This is owed to both MTP moment and angle being zero at the beginning of EP, regardless of condition. In fact, this is almost guaranteed by the methodology and anatomy—the foot is nearly flat on the ground (angle = 0) when the center of pressure crosses the MTP joint center (moment = 0) (Figures 3 and 4). This was different than the LTR phase, which showed an increase in constant terms that were significantly different for each speed ($P < .001$). However, this merely reflects the increase in peak moment with walking speed since the regression's intercept needs to be greater to account for the higher peak moment.

Discussion

This study was the first, to our knowledge, to quantify dynamic MTP joint stiffness during the toe-rocker phase of stance across a

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Table 1 Stiffness Values, Regression Constants, Peak MTP Joint Angle, Peak GRF, Peak MTP Joint Moment, and Regressions Statistics (r^2 and RMSE)

	0.4 statures·s^{-1}	0.6 statures·s^{-1}	0.8 statures·s^{-1}	1.0 statures·s^{-1}
ETR stiffness, N·m·deg^{-1}·body weight^{-1} ^	0.00070	0.00079	0.00091	0.00109
SD	0.00037	0.00033	0.00035	0.00038
P value	.066	.014*	<.001*	
LTR stiffness, N·m·deg^{-1}·body weight^{-1} ^	−0.00026	−0.00036	−0.00042	−0.00051
SD	0.00013	0.00016	0.00017	0.00019
P value	<.001*	.002*	<.001*	
ETR constant, N·m·deg^{-1}·body weight^{-1} ^	−0.00077	−0.00112	−0.00189	−0.00325
SD	0.00214	0.00304	0.00307	0.00335
P value	.756	.434	.081	
LTR constant, N·m·deg^{-1}·body weight^{-1} ^	0.01487	0.02116	0.02744	0.03636
SD	0.00655	0.00842	0.00961	0.01087
P value	<.001*	<.001*	<.001*	
Peak MTP joint angle, deg	53.90	58.26	64.70	70.37
SD	7.37	6.18	6.25	6.46
P value	<.001*	<.001*	<.001*	
Peak normalized GRF, N·body weight^{-1} ^	0.934	1.057	1.151	1.207
SD	0.155	0.052	0.058	0.080
P value	<.001*	<.001*	<.001*	
Max MTP moment, N·m·body weight^{-1} ^	0.009	0.012	0.016	0.020
SD	0.004	0.005	0.006	0.006
P value	<.001*	<.001*	<.001*	
r^2—ETR	.946	.953	.955	.961
SD	0.025	0.016	0.014	0.012
r^2—LTR	−.989	−.992	−.993	−.995
SD	0.015	0.006	0.007	0.003
RMSE—ETR	1.17E-06	1.87E-06	2.86E-06	4.49E-06
RMSE—LTR	2.10E-07	4.68E-07	6.06E-07	8.58E-07

Abbreviations: ETR, early toe rocker; GRF, normalized ground reaction force; LTR, late toe rocker; MTP, metatarsophalangeal; RMSE, root mean squared error. *Conditions are significantly different ($P < .05$) based on MLM results. ^Dominant/nondominant limbs significantly different ($P < .05$) between all conditions, controlling for subject.

range of controlled *walking* speeds, and based off of stance phases that correspond to MTP joint mechanics. MTP stiffness was separated into 2 subphases,—ETR and LTR. At every walking speed, ETR stiffness magnitude was higher than LTR stiffness magnitude by a factor of 2 to 3. MTP joint stiffness increased with walking speed, much like ankle dynamic stiffness,[37] and MLMs indicated most of those changes were statistically significant. These findings generally supported our hypothesis that MTP stiffness would increase with walking speed and follow the pattern of results that indicate MTP stiffness increases during running.[24,38] This study provides a more comprehensive characterization of MTP joint stiffness that will be critical for advancing the design of orthotic and prosthetic foot and ankle–foot devices, and associated user outcomes. Clinically, if a patient exhibits low MTP stiffness due to an injury or impairment, and an orthotic foot plate's stiffness could be tuned to match the difference between that patient's MTP stiffness and typical healthy MTP stiffness, as Arch and Reisman[6] have done regarding ankle joint stiffness and ankle–foot orthosis stiffness. This study shows that MTP stiffness changes with speed, so any orthotic foot plate tuned to an individual should also take that individual's comfortable walking speed into account.

The regression coefficient k was positive for ETR and negative for LTR. A positive regression coefficient represents a true stiffness, during which moment and angle increase (or decrease) simultaneously. A negative regression coefficient does not represent a realistic stiffness since moment and angle are not changing in the same positive or negative direction. This is similar to what Davis and DeLuca[1] described as the "active facilitation" phase of stance regarding ankle stiffness, during which ankle angle begins to decrease but ankle moment continues to increase between the loading and push-off phases of stance. In the case of the MTP joint, the MTP angle and moment are changing in different directions, but they are opposite of the ankle's active facilitation phase—here the moment is decreasing due to the decreasing GRF, but the angle continues to increase as the leg progresses forward. In either case, no simple passive material can behave that way. Thus, LTR "stiffness" is not a stiffness at all, but a more complex dynamic motion. Additionally, limb dominance was shown to have a statistically significant effect on some of our outcome measures (Table 1). However, this significance did not elucidate any specific trends. For example, the dominant limb had higher average ETR stiffness values for the slow and fast speeds, but not

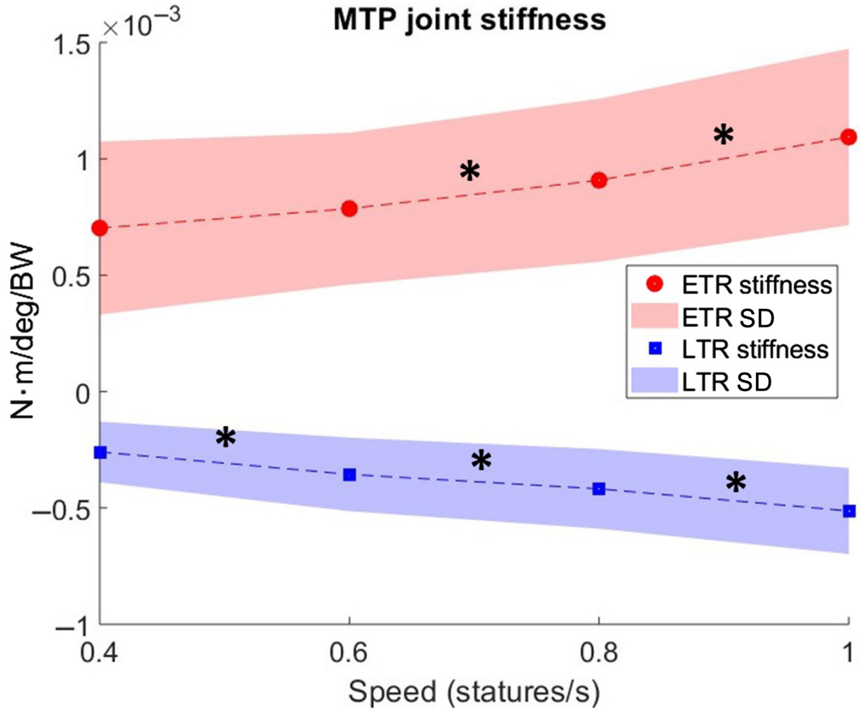

Figure 4 — Average MTP joint stiffness across walking speeds. Loading phase (circles) stiffness has a greater magnitude and greater SD than the push-off phase (squares) stiffness. Asterisks above/below lines in between points denote a significant difference between the 2 adjacent walking speeds. BW indicates body weight; ETR, early toe rocker; LTR, late toe rocker; MTP, metatarsophalangeal.

the medium speed. Likewise, no clear trend was exhibited between limbs for LTR stiffness, or peak kinetic, and kinematic values.

ETR stiffness is important to orthotic and prosthetic device design that aims to provide facilitation or imitation of natural human motion. While imitation of the MTP joint is not necessarily the goal of an orthotic device, being able to provide support that broadly follows the natural patterns of gait will ensure that the user is not “fighting” against the device with their existing limb in order to achieve proper kinematics, as suggested by Arch and Reisman.[6] During toe rocker, maintaining those kinematics means allowing the MTP joint to progress through a wide angle range. Prosthetics, on the other hand, typically aim to imitate and replace the entire function of a given body part. This imitation cannot be achieved by passive devices alone, since no musculature exists within the device to generate energy, and due to LTR’s negative stiffness. However, passive elements used in conjunction with active elements may provide robust energy storage and return characteristics that benefit users. Understanding the fundamental biomechanics may provide design insight for hybrid active–passive prostheses such as the Ossür Proprio (Ossür hf.) and the Ottobock Empower (Ottobock).

There were 2 notable assumptions for our model: (1) The MTP joint center was assumed to be directly on the surface of the force plate, and (2) the toes were assumed to be flat on the ground for the duration of terminal stance. Because of these 2 assumptions, any mediolateral GRF and anterioposterior GRF act directly through, not about, the MTP joint center and thus do not contribute to the net moment about it for the entirety of terminal stance. Additionally, the anterioposterior GRF has a negligible contribution to the overall GRF during terminal stance. Including the anterioposterior GRF in the overall GRF calculation resulted in less than a 3% increase in GRF magnitude compared with only the vertical GRF across all conditions. As such, only vertical GRF was used to calculate the MTP joint moment. An additional limitation of our methodology was that subjects were not screened for MTP joint range of motion in order to obtain data from as broad a range of healthy individuals as possible. A subject with limited range of motion at the MTP joint may increase stiffness values as long as that subject generates a similar peak moment about the MTP joint. However, other gait adaptations associated with limited MTP range of motion that might affect net MTP stiffness are possible. This study does not imply that full or partial-length foot orthoses are preferred, but rather implies that the net stiffness properties of such a device about the MTP joint are the more important design criterion.

The most important limitation of this study is that it observed young, healthy individuals, which would not typically use orthotic or prosthetic foot and ankle–foot devices. However, the goal of this study is not necessarily to characterize how patients’ MTP stiffness is affected by walking speed, but to map the range of healthy MTP stiffness to which future orthotic and prosthetic devices could be designed. If only the MTP stiffness properties of patient populations are known, then there would be no baseline for typical MTP stiffness values that we would want those patients to work toward for, say, a muscle-strengthening protocol or for a customized foot orthosis designed to sum with a patient’s existing MTP stiffness to match a typical healthy stiffness value. The characteristics of both healthy and patient populations must be known in order to quantify *how much* a patient is impaired with respect to a healthy population. Future work can use the stiffness values obtained from this study to act as goal targets for patient populations.

The MTP stiffness values reported in this paper broadly coincide with Sanchis-Sales et al[23] and Farris et al,[24] though reported in different units. However, critically, methodologies differ between those studies and this study for defining phases of stance and calculating joint moments and angles, which makes direct comparisons difficult. Direct stiffness value comparisons to the study from Mager et al[39] is easier, though that study only examined a single walking and jogging speed, and only examined 1 region of stance (approximately corresponding to our ETR phase). Mager et al[39] reported MTP stiffness values of $1.2e^{-3}$ N·m·deg^{-1}·body weight^{-1}, while our ETR stiffness values at 0.8 statures·s^{-1} were only slightly lower, at $0.9e^{-3}$ N·m·deg^{-1}·body weight^{-1}. Finally, an advantage of this study compared with those previous studies is that a relatively simple marker set was used. While a higher degree of specificity could be obtained from multi-segment foot-specific marker sets,[20,21,39] the marker set, and single-segment foot model used in this study provide an advantage in simplicity. Regardless, our stiffness values were still quite close to the more complex model of Mager et al.[39] This might mean that the more complex marker sets may not be necessary compared with our relatively simple one.

This study provides a comprehensive baseline for MTP stiffness during toe-rocker subphases throughout a range of slow and fast walking speeds in young, healthy individuals. Mechanically, the MTP joint does not act like a perfect spring during this entire phase of stance, but does during EP. A more dynamically complex motion occurs during LP. This difference in EP stiffness versus LP moment-angle relationships stiffness may prove important in optimizing future orthotic and prosthetic designs that better mimic, facilitate, and support natural human walking.

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