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Original Article

Analysis of windlass mechanism according to one walking cycle

Takashi Shiroshita, PT, PhD[1)]

[1)] School of Physical Therapy, Faculty of Rehabilitation, Gunma PAZ University: 1-7-1 Tonyamachi, Takasaki-shi, Gunma 370-0006, Japan

Abstract. [Purpose] This study aimed to calculate the windlass mechanism in one walking cycle (WC) using the medial longitudinal arch (MLA) height and compare its mechanism with joint moments, angles, and center of gravity movement. [Participants and Methods] The study analyzed the gait of 20 healthy adults (14 males, six females) using a three-dimensional motion analyzer to calculate several parameters. [Results] In the terminal stance, the MLA height reached 20.6 ± 6.0 mm (minimum value) at 49% WC. Simultaneously, the ankle dorsiflexion angle, ankle internal plantarflexion moment, and forward shift of the center of gravity reached the maximum values. At 62% WC, the MLA height was 26.8 ± 4.8 mm and reached maximum during the stance phase, indicating a windlass mechanism. Additionally, the MLA height was 61.7 ± 22.7 mm at 69% WC, indicating an MLA spiking phenomenon. [Conclusion] The MLA height was lowest at 49% WC due to reverse windlass mechanism. Although the windlass mechanism was activated at 62% WC, it was functionally equivalent to the swing phase. Push-off was impossible during the swing phase. At 69% WC, the swing phase showed a second windlass mechanism.
Key words: Windlass mechanism, Reverse windlass, Push off

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INTRODUCTION

Anatomically, the medial longitudinal arch (MLA) is supported by the calcaneal, talus, navicular, medial cuneiform, and three medial metatarsal bones. The plantar fascia, calcaneonavicular (spring) ligament, tibialis posterior, medial tarsometatarsal joint, and extrinsic and intrinsic muscles of the foot all assist in maintaining the height and shape of the MLA[1–7)]. Passive factors (bones and ligaments), including the plantar fascia, have been traditionally considered to be vital to the MLA[1)]. However, several reports have claimed that active factors support the MLA. Researchers who support active factors are divided into two groups: those who support extrinsic muscles[8, 9)] and those who support intrinsic muscles[2–4)]. In recent years, short foot exercises are a typical example that is performed in intrinsic muscle support groups. Furthermore, the intrinsic muscles of the foot have been reported as an important factor in the MLA[6, 7, 10, 11)]. We have also described a lesser-toe exercise (LTE) we have designed[12–14)]. The toe position for the LTE is the distal interphalangeal joint extension and proximal interphalangeal joint flexion positions, similar to that for the short foot exercise[6, 7–10, 11)]. The LTE may strengthen the intrinsic muscles; however, it does not use the great toe, as only the four other toes and ankle are held in a plantar flexion position.

In 1954, Hicks provided the first description of the windlass mechanism, stating that it is induced by the proximal pressure exerted by the proximal phalanx on the corresponding metatarsal ray[15)]. The windlass mechanism is not the only mechanism that supports the MLA; beam and truss mechanisms also do. The beam mechanism is generated by the action of the bones, joints, and ligaments[16)]. Alternatively, the truss mechanism supports the MLA through the plantar aponeurosis, which extends from the calcaneus tubercle to the metacarpophalangeal (MP) joint, the tendon sheath of the flexor tendon, and the base of each proximal phalanx. The functional characteristics of foot weight-bearing are determined by the interaction between the

Corresponding author. Takashi Shiroshita (E-mail: t-shiroshita@nifty.com)

beam and truss mechanisms. For example, if the beam mechanism acts excessively, and the truss mechanism deteriorates, problems such as calluses may develop[16–18].

Thus, classifying the windlass mechanism as a passive or active factor is difficult. While dorsiflexion of the first row is active, no active factor is associated with the plantar aponeurosis, and it is thus passive. Based on these contradicting mechanisms, we consider the windlass mechanism as a passive factor that is associated with an active factor, as it depends on the interaction between active and passive factors.

Multiple reports have investigated the walking cycle (WC)[19–22], but there are different interpretations of WC. The present study uses the most well-known definition proposed by Perry et al[22, 23]. They classified the stance phase as follows: Loading response (LR) is defined as the time from the initial contact (IC) to approximately 12% WC; the mid-stance (MS), at approximately 12%–31% WC; the terminal stance (TS), approximately 31%–50% WC; and the pre-swing (PS), approximately 50%–62% WC[23]. Naturally, swing phase is defined as 62%–100%.

In the late stance, the MP joint extends (column dorsiflexion) while the heel is raised, and the plantar aponeurosis tenses. The MLA is believed to increase as the windlass mechanism occurs, causing a push-off phenomenon and propulsion motion[15, 24]. During TS (31%–50% WC), the soleus and gastrocnemius muscles are stretched and tensed by eccentric contraction. Subsequently, at PS (50%–62% WC), these muscles loosen rapidly, generating force. This force is called recoil due to the catapult action. The propulsion motion caused by the catapult action is described as a push-off phenomenon[23]. This push-off phenomenon could not be due to muscle activity alone[25–27]. In addition, the opposite foot has a shock-absorbing effect from the IC to the LR. The windlass mechanism could also be related to the push-off phenomenon: As the plantar aponeurosis tenses, the windlass rolls up, and the ankle joint power reaches its peak value during the PS[15, 28].

However, recent reports have described that even though the MP joint continues to dorsiflex during the late stance, the plantar aponeurosis does not tense. These studies argue that the windlass mechanism is not caused by tension in the plantar aponeurosis but by other factors, such as the foot intrinsic muscles[29–31]. As mentioned above, many recent reports assert that foot intrinsic muscles are important in MLA. Moreover, the detailed period of late stance was unclear. It was found that the windlass mechanism occurs during the “late stance”' period, which is about 50% of the WC when converting data from several previous studies into a WC[24, 29–31]. In this definition, 50% WC is the TS rather than the PS; the ankle joint moment is at its peak value; the power is close to 0; and no push-off phenomenon has yet occurred.

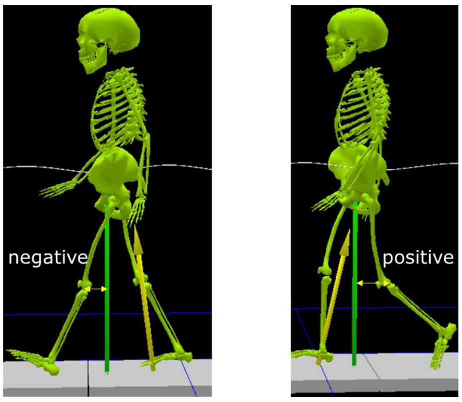
Various contradictory interpretations exist, and it is still debatable as to which interpretation is valid. The primary aim of this study was to clarify those contradictions and questions by calculating the MLA height (as the visual target of windlass mechanism) during walking and measuring it in synchronization with the ground reaction force, joint moment, joint angle, and movement of the center of gravity.

PARTICIPANTS AND METHODS

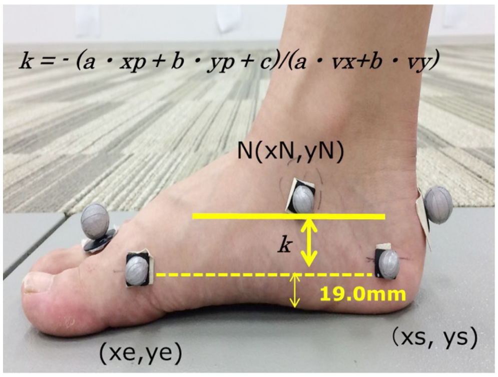
Twenty participants (14 males and 6 females) who had not been admitted to a medical institution due to injury around the ankle for the past 6 months were enrolled. The mean ± standard deviation (SD) of the age of participants was 21.2 ± 0.4 years; the mean ± SD height was 166.4 ± 6.6 cm; and the mean ± SD weight was 59.4 ± 6.1 kg. The research was fully explained to all participants, and measurements were conducted with their consent. This study was conducted in line with the guidelines of the Ethics Committee for Human Research of Gunma Paz University (approval number: PAZ14-22).

This is a cross-sectional study designed using three-dimensional motion analysis. A three-dimensional motion analyzer Vicon MX (Vicon Motion Systems, Oxford, UK), nine infrared cameras (T10 Vicon Motion System, Nexus 1.8.5), and three-floor reaction force meters (AMTI: ADVANCED MECHANICAL TECHNOLOGY INC. Watertown, MA, USA) were used as the measurement instrument. This study adopted a full plug-in model. The sampling frequency of the camera and ground reaction force was 100 Hz. The starting position for measurement was 3.5 m away from the floor reaction force meter, and the ending position was also 3.5 m away from the floor reaction force meter. Measurements were performed by asking the participants to free walk three times, and the average values were calculated. One WC was normalized to 100%, and the ankle internal plantar flexion moment, ankle dorsiflexion angle, and center of gravity movement were measured. The values at 12%, 31%, and 50% WC were determined. Then, the amount of change from 12%–31% WC for the MS and that from 31%–50% WC for the TS were calculated and compared. The center of gravity movement was calculated as the distance from the marker of the knee joint axis to the center of gravity line. The value was negative and positive when the center of gravity was behind the knee axis, in front of the knee axis, respectively. Therefore, a positive number means more anterior to the knee axis, a high value means more forward, and a low value means more backward (Fig. 1a). Statistical analysis was performed by comparing each change using the t-test performed with IBM SPSS Statistics 21 (IBM Corp., Armonk, NY, USA).

The MLA height was calculated by adding three more markers to the full plug-in model described above. These markers were first applied to the calcaneus and first metatarsal at a height of 19 mm from the floor, and then to the navicular bone. The MLA height was calculated as the distance between the calcaneus and navicular bone with the first metatarsal as the base[32–34] (Fig. 1b). Furthermore, in this study, the analysis and consideration of LR from 0% to 12% was deliberately omitted because the opposite side was PS and the focus was on the windlass mechanism.



a) Center of gravity movement



b) Medial longitudinal arch (MLA) height

Fig. 1. Three-dimensional motion analyzer a) The calculation method is the distance from the marker of the knee joint axis to the line of center of gravity. When the center of gravity is behind knee axis, the value is negative; however, when the center of gravity line is in front of knee axis, it is set as a positive value. b) Medial longitudinal arch (MLA) height. The calculation method is the distance between the calcaneus and navicular bone with the first metatarsal as the bas. The calculation formula is as follows: **k= −(a • xp + b • yp + c)/(a • vx+b • vy) k=MLA height, a=ye −ys b=xs −xe c=−(a • xs + b • ys) L=√((xe−xs)2 + (ye−ys)2) vx=−ey=−(ye −ys) / L, vy=ex=(xe −xs) / L**

RESULTS

The mean values at 12%, 31%, and 50% WC of ankle internal plantar flexion moments in the ankle joint were 0.14 ± 0.14 Nm, 0.61 ± 0.17 Nm, and 1.32 ± 1.13 Nm, respectively. The amount of moment change value in MS was 0.47 ± 0.16 Nm, while the TS value was 0.71 ± 0.20 Nm. The amount of moment change value in TS was significantly higher than that of the MS value (p=0.000; Table 1, Fig. 2). Conversely, the mean ± SD ankle dorsiflexion angles at 12%, 31%, and 50% WC were 1.1° ± 2.8°, 8.5° ± 3.9°, and 12.3° ± 5.6°, respectively. The amount of dorsiflexion angle change value in MS was 7.3° ± 3.3°. while the TS value was 3.8° ± 2.6°. The angle amount of change value in MS was significantly higher than that of the TS (p=0.000; Table 1, Fig. 2). The mean ± SD gravity movement at 12%, 31%, and 50% WC were −114.4 ± 27.8 mm, 72.5 ± 19.7 mm, and 183.0 ± 15.9 mm, respectively. The amount of gravity movement change value in MS was 186.9 ± 20.4 mm, while that of the TS was 110.5 ± 20.0 mm. The amount of gravity movement change value in MS was significantly higher than that of the TS (p=0.000; Table 1, Fig. 2). The MLA height was at the minimum at 49% WC (20.6 ± 6.0 mm) and at the maximum during the stance phase at approximately 62% WC (26.8 ± 4.8 mm). Furthermore, 62% WC was the endpoint of the stance phase, or the start point of the swing phase. The MLA spike phenomenon was observed at 69% WC (61.7 ± 22.7 mm), 69%WC is in the swing phase. This phenomenon has not been reported previously. Interestingly, many participants showed the spike phenomenon at exactly 69% WC (Fig. 2). Furthermore, to clarify what occurred at each % WC, a combined graph of the vertical components of ground reaction force, joint angle, moment, center of gravity movement, and MLA height was constructed (Fig. 2). At 50% WC, the maximum ankle dorsiflexion angle (12.3°), maximum ankle internal plantarflexion moment (1.32 Nm), and maximum forward movement of the center of gravity (183.0 mm) were synchronized. However, the MLA height was at its minimum value.

Table 1. Comparison between MS and TS

	AM (Nm)**	AA (°)**	GM (mm)**
MS	0.47 ± 0.16	7.3 ± 3.3	186.9 ± 20.4
TS	0.7 ± 0.20	3.8 ± 2.6	110.5 ± 20.0

Mean ± SD. ** $p<0.01$.
MS: mid stance; TS: terminal stance; AM: ankle internal plantar flexion moment (Nm); AA: ankle angle (°); GM: gravity movement (mm). The internal plantar flexion moment, the TS mean value was significantly high than MS (N=20, p=0.000 <0.01). The ankle angles, the MS mean value was significantly higher than TS (N=20, p=0.000 <0.01). The gravity movement, the MS mean value was significantly higher than TS (N=20, p=0.000 <0.01). Statistical analysis was performed by comparing each change amount with a t-test, and analysis software used was IBM SPSS Statistics 21.

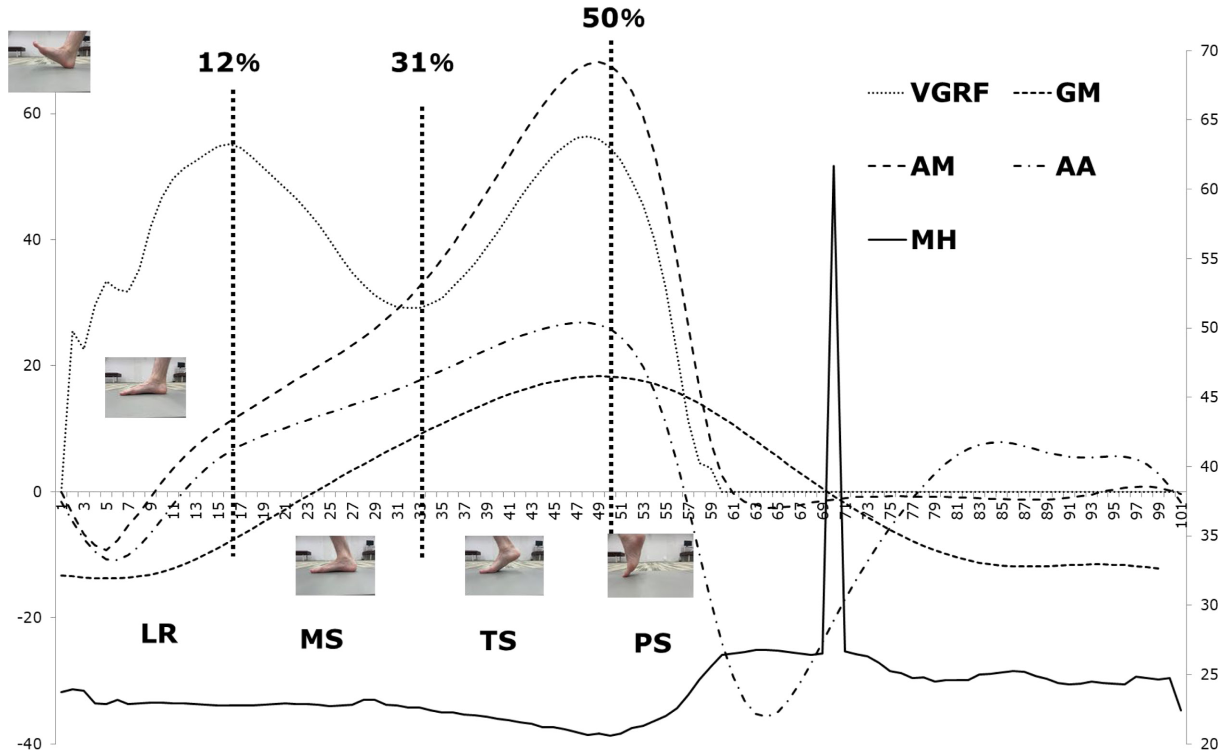


Fig. 2. Comparison of medial longitudinal arch (MLA) based on various parameters.
VGRF: vertical component of ground reaction force; GM: gravity movement (mm); AM: Ankle moment (Nm); AA: Ankle angle (°); MH: MLA height (mm); LR: loading response; MS: midstance; TS: terminal stance; PS: pre-swing; WC: walking cycle (%).

DISCUSSION

The present study used three-dimensional motion analysis to explore the windlass mechanism by calculating the MLA height for each step cycle. In MS at 12%–31% WC, both the ankle dorsiflexion angle and the moment showed a linear waveform rising to the right. The ankle dorsiflexion angle in the MS was bigger than that in the TS. The forward gravity movement distance was higher in the MS than in the TS. MS appears to be a phase in which the center of gravity moves forward due to joint movement. Conversely, the MLA height hardly changed, and the windlass mechanism was not activated in this phase. Similar to MS, in TS at 31%–50% WC, both the ankle dorsiflexion angle and moment showed a linear waveform rising to the right. The ankle joint moment in the TS was superior to that in the MS. Notably, the minimum value of the MLA was 20.7 ± 6.0 mm at 49% WC, seemingly due to the reverse windlass mechanism in TS. In PS at 50%–62%WC, MLA height reached its maximum value during the stance phase. In other words, the occurrence of the windlass mechanism in PS is similar to previous studies[29–31], but the results of the current study showed that WC was 62% even in PS. At 62% WC, the windlass mechanism was demonstrated with almost no toe contact. The results of this study differed from the cycle reported in previous studies[29, 30]. At 69% WC in the swing phase, a spike phenomenon appeared that resembled the windlass mechanism but was approximately three times larger than that in the stance phase. This phenomenon may have a different kinematic role from the windlass mechanism what has been known to date.

Several researchers have argued that propulsion and push-off phenomena are negative in the TS because tibial advancement slows down in TS[35, 36]. Furthermore, the center of gravity movement was lower in the TS than the MS. The TS is a phase of single-leg support that uses instability as a source of power. To safety land, the eccentric contraction of the triceps surae muscle decelerates the anterior tilt of the tibia to safely, land, and the ankle moment appears to facilitate a progression movement while landing safely. Therefore, the TS is a phase in which control is performed to prevent the center of gravity from dropping suddenly due to the unstable situation of one-leg support.

Several studies[29–31] have reported that the windlass mechanism occurs during the "late stance". The present study found that it occurred at 62% WC. As in previous studies, a windlass mechanism occurred in the PS during the late stance. In previous studies, only the stance phase was measured without measuring the one WC, and when converted to one WC, Windlass mechanism occurred at about 48% WC in Carava et al.[29] and about 46.5% WC in Fess et al[30]. Carava et al. measured the MLA angle, whereas Fess et al. measured the length of the sole. Thus, they actually measured the windlass mechanism at the MLA height as in the current study, not angle or length.

In the MS, an attention should be paid to around 50% WC, which is the endpoint of the TS. During this phase, ankle dorsiflexion, internal plantar flexion moment, and anterior shift of the center of gravity all reach their peak values. Conversely, the MLA height is at the minimum at around 50% WC (Fig. 2). This phenomenon of decreased MLA occurs when the load is applied to the arch of the foot, which Hicks designated as the reverse windlass mechanism[15, 24]. This reverse windlass mechanism might be also a truss mechanism.

In the PS, several studies[29–31] have considered that the windlass mechanism during the late stance is due to the intrinsic muscles of the foot, rather than the tension of the plantar aponeurosis caused by the truss. However, when the foot's intrinsic muscles are activated, the toes must be in contact with the ground. As shown in Fig. 2, while the windlass mechanism occurs in the PS, at 62% WC of cases, the toes are hardly in contact with the floor; thus, functionally, the occurrence is during the swing phase rather than the stance phase. The windlass mechanism occurs during the functional swing phase. Thus, the relationship between the push-off and windlass mechanism would be difficult to establish. When the toes make contact during the PS, the windlass mechanism needs to occur at approximately 50%–55% WC in the first half of the PS[4, 37–39]. Furthermore, the PS occurs in the double-leg support phase, and safe weight transfer to the opposite side (LS: 0–12%) is an important function. The windlass mechanism also functions to stably shift the weight to the opposite side. In the latter half of the PS, the support phase base rapidly decreases from the forefoot to the toes. In other words, the windlass mechanism is generated to obtain stability during a phase of kinematic instability. If this action is not due to the windlass mechanism activated by the plantar aponeurosis, then it would be due to the beam mechanism. In other words, the MLA height increased due to the beam mechanism.

The windlass mechanism-like spike phenomenon that appeared in the swing phase was larger than that in the stance phase. This second windlass mechanism is different from the well-described windlass mechanism in that it may have a kinematic role. Although the second windlass phenomenon did not occur in five participants, it occurred at 69% WC consistently among those who expressed it. Furthermore, 69% WC is the early swing phase, when the ankle transits from plantarflexion to dorsiflexion. The ankle joint is at the mid-position at approximately 76% WC (Fig. 2), while at 69% WC, it is still in the plantar flexion position. The spike phenomenon is considered as an initial reaction for the toes to change from flexion to extension and for the ankle to obtain dorsiflexion.

The conclusions of this study are that the reverse windlass mechanism in the TS is likely the trigger for the first windlass mechanism by the beam mechanism, and its function is not a push-off but a stabilization mechanism. Conversely, the second windlass mechanism may be caused by the toes automatically extending during the swing phase and the tensing of the plantar aponeurosis. There may be a second windlass mechanism that makes it easier to swing out.

Conflict of interest

Authors declare no conflicts of interest associated with this manuscript.

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REFERENCES

1) Basmajian JV, Stecko G: The role of muscles in arch support of the foot. J Bone Joint Surg Am, 1963, 45: 1184–1190. [Medline] [CrossRef]

2) Fiolkowski P, Brunt D, Bishop M, et al.: Intrinsic pedal musculature support of the medial longitudinal arch: an electromyography study. J Foot Ankle Surg, 2003, 42: 327–333. [Medline] [CrossRef]

3) Headlee DL, Leonard JL, Hart JM, et al.: Fatigue of the plantar intrinsic foot muscles increases navicular drop. J Electromyogr Kinesiol, 2008, 18: 420–425. [Medline] [CrossRef]

4) Jam B: Evaluation and retraining of the intrinsic foot muscles for pain syndromes related to abnormal control of pronation. Advanceed Physical Therapy Education Institute, 2006. https://www.aptei.ca/wp-content/uploads/Intrinsic-Muscles-of-the-Foot-Retraining-Jan-29-05.pdf (Accessed Jan. 2022)

5) Jones RL: The human foot: an experimental study of its mechanics, and the role of its muscle and ligaments in the support. Am J Anat, 1941, 68: 1–39. [CrossRef]
6) Jung DY, Kim MH, Koh EK, et al.: A comparison in the muscle activity of the abductor hallucis and the medial longitudinal arch angle during toe curl and short foot exercises. Phys Ther Sport, 2011, 12: 30–35. [Medline] [CrossRef]
7) Jung DY, Koh EK, Kwon OY: Effect of foot orthoses and short-foot exercise on the cross-sectional area of the abductor hallucis muscle in subjects with pes planus: a randomized controlled trial. J Back Musculoskeletal Rehabil, 2011, 24: 225–231. [Medline] [CrossRef]
8) Narváez J, Narváez JA, Sánchez-Márquez A, et al.: Posterior tibial tendon dysfunction as a cause of acquired flatfoot in the adult: value of magnetic resonance imaging. Br J Rheumatol, 1997, 36: 136–139. [Medline] [CrossRef]
9) Funk DA, Cass JR, Johnson KA: Acquired adult flat foot secondary to posterior tibial-tendon pathology. J Bone Joint Surg Am, 1986, 68: 95–102. [Medline] [CrossRef]
10) Lynn SK, Padilla RA, Tsang KK: Differences in static- and dynamic-balance task performance after 4 weeks of intrinsic-foot-muscle training: the short-foot exercise versus the towel-curl exercise. J Sport Rehabil, 2012, 21: 327–333. [Medline] [CrossRef]
11) Mulligan EP, Cook PG: Effect of plantar intrinsic muscle training on medial longitudinal arch morphology and dynamic function. Man Ther, 2013, 18: 425–430. [Medline] [CrossRef]
12) Shiroshita T, Fukubayashi T: Surface electromyogram analysis of toe exercises: a comparison of toe function. J Phys Ther Sci, 2012, 24: 59–62. [CrossRef]
13) Shiroshita T, Fukubayashi T: Comparison of towel-gathering exercise and toe exercises for the painful accessory navicular. J Phys Ther Sci, 2011, 23: 455–458. [CrossRef]
14) Shiroshita T: Morphologic relationship between toe exercises and the medial longitudinal arch. J Nov Physiother, 2019, 9: 1000420.
15) Hicks JH: The mechanics of the foot II. The plantar aponeurosis and the arch. J Anat, 1954, 88: 25–30. [Medline]
16) Hicks JH: The foot as a support. Acta Anat (Basel), 1955, 25: 34–45. [Medline] [CrossRef]
17) Lapidus PW: Kinesiology and mechanical anatomy of the tarsal joints. Clin Orthop Relat Res, 1963, 30: 20–36. [Medline] [CrossRef]
18) Sarrafian SK: Functional characteristics of the foot and plantar aponeurosis under tibiotalar loading. Foot Ankle, 1987, 8: 4–18. [Medline] [CrossRef]
19) Murray MP, Kory RC, Clarkson BH, et al.: Comparison of free and fast speed walking patterns of normal men. Am J Phys Med, 1966, 45: 8–23. [Medline] [CrossRef]
20) Inman VT: Human locomotion. Can Med Assoc J, 1966, 94: 1047–1054. [Medline]
21) Seibel MO: Foot Function: a programmed text. In: Chapter 4. Baltimore: Williams & Wilkins, 1988, pp 41–50.
22) Jacquelin Perry M: Gait analysis normal and pathological function. In: Section 1 fundamentals. Thorofare: SLACK, 1992, pp 1–47.
23) Jacquelin Perry M: Gait analysis normal and pathological function, 2nd ed. In: Section 1 Fundamentals. Thorofare: SLACK, 2010, pp 1–47.
24) Farris DJ, Birch J, Kelly L: Foot stiffening during the push-off phase of human walking is linked to active muscle contraction, and not the windlass mechanism. J R Soc Interface, 2020, 17: 20200208. [Medline] [CrossRef]
25) Alexander RM, Bennet-Clark HC: Storage of elastic strain energy in muscle and other tissues. Nature, 1977, 265: 114–117. [Medline] [CrossRef]
26) Ishikawa M, Komi PV, Grey MJ, et al.: Muscle-tendon interaction and elastic energy usage in human walking. J Appl Physiol, 2005, 99: 603–608. [Medline] [CrossRef]
27) Maganaris CN, Paul JP: Tensile properties of the *in vivo* human gastrocnemius tendon. J Biomech, 2002, 35: 1639–1646. [Medline] [CrossRef]
28) Stolwijk NM, Koenraadt KL, Louwerens JW, et al.: Foot lengthening and shortening during gait: a parameter to investigate foot function? Gait Posture, 2014, 39: 773–777. [Medline] [CrossRef]
29) Caravaggi P, Pataky T, Günther M, et al.: Dynamics of longitudinal arch support in relation to walking speed: contribution of the plantar aponeurosis. J Anat, 2010, 217: 254–261. [Medline] [CrossRef]
30) Fessel G, Jacob HA, Wyss C, et al.: Changes in length of the plantar aponeurosis during the stance phase of gait—an *in vivo* dynamic fluoroscopic study. Ann Anat, 2014, 196: 471–478. [Medline] [CrossRef]
31) Sichting F, Ebrecht F: The rise of the longitudinal arch when sitting, standing, and walking: contributions of the windlass mechanism. PLoS One, 2021, 16: e0249965. [Medline] [CrossRef]
32) Nielsen RG, Rathleff MS, Simonsen OH, et al.: Determination of normal values for navicular drop during walking: a new model correcting for foot length and gender. J Foot Ankle Res, 2009, 2: 12. [Medline] [CrossRef]
33) Rathleff MS, Nielsen RG, Kersting UG: Navicula drop test ad modum Brody: does it show how the foot moves under dynamic conditions? J Am Podiatr Med Assoc, 2012, 102: 34–38. [Medline] [CrossRef]
34) Christensen BH, Andersen KS, Pedersen KS, et al.: Reliability and concurrent validity of a novel method allowing for in-shoe measurement of navicular drop. J Foot Ankle Res, 2014, 7: 12. [Medline] [CrossRef]
35) Gilbert JA, Maxwell GM, McElhaney JH, et al.: A system to measure the forces and moments at the knee and hip during level walking. J Orthop Res, 1984, 2: 281–288. [Medline] [CrossRef]
36) Simon SR, Mann RA, Hagy JL, et al.: Role of the posterior calf muscles in normal gait. J Bone Joint Surg Am, 1978, 60: 465–472. [Medline] [CrossRef]
37) Rothermel SA, Hale SA, Hertel J, et al.: Effect of active foot positioning on the outcome of a balance training program. Phys Ther Sport, 2004, 5: 98–103. [CrossRef]
38) Campbell EB, Frye JL, Gribble PA: Strengthening of the plantar intrinsic foot muscles decreases navicular drop and decreases muscular fatigue. J Athl Train, 2008, 43: S123.
39) Sauer LD, Beazell J, Hertel J: Considering the intrinsic foot musculature in evaluation and rehabilitation for lower extremity injuries: a case review. Athl Train Sports Health Care, 2011, 3: 43–47. [CrossRef]