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On the Mechanics of the Knee during the Stance Phase of the Gait

Kamran Shamaei and Aaron M. Dollar, *Member, IEEE*

***Abstract*— In this paper, we explore the mechanical behavior of the knee during the weight acceptance stage of stance during normal walking. We show that the torque/angle behavior of the knee during this stage can be approximated by a linear torsional spring. The mechanical parameters completely specifying this spring are identified, including stiffness, amount of rotation, and angle of engagement, and the effect of gait speed and body/load mass on those parameters are discussed. We discuss how the findings of this paper can be applied to the design of leg orthoses, prostheses and exoskeletons, and bipedal robots in general, allowing the implementation of human-like leg compliance during stance with a relatively simple latched-spring mechanism.**

I. Introduction

THERE are numerous benefits to being able to mimic human-like gait mechanics with simple engineered mechanisms. These range from anthropomorphic bipedal robots ([1] and [2]) to leg exoskeletons and orthoses ([3] and [4]) and prostheses. This paper represents a first step along the path of the development of a novel stance-control orthosis that does not rigidly lock the knee during stance, but instead implements a natural amount of knee compliance during the weight acceptance phase.

There are several mechanisms that can lead to a pathological gait, including muscular weakness, pain, deformity, muscular spasticity and sensory loss [5]. Depending on the joint and phase of the gait, these mechanisms can result in inadequate/excessive flexion/extension [5]. Treatment of these dysfunctions is usually carried out using a passive orthosis to support, limit range of motion, stabilize, and/or assist the knee. These devices are often used to treat brain-injured patients, spinal cord injured patients, patients with lower extremity fractures, patients with sports-related disorders, post-polio syndrome, and neuropathic and dysvascular patients [6].

The knee has a wide range of roles during the execution of gait, including supporting the body weight and deceleration during the stance phase (by applying a large moment [7]). Thousands of patients suffer knee joint disorders, such as quadriceps weakness and pain, resulting in a pathological stance phase. Quadriceps weakness can result in collapse or hyperextension of the knee from the onset of the stance phase until mid-stance. Patellofemoral pain syndrome, the reason for 20% to 40% of visits to sport medicine clinics [8], also causes a pathological gait. Other than these two examples, thousands suffer from injuries, stroke, post-polio, multiple sclerosis, SCI, etc. An assistive orthosis capable of unloading the knee in the stance phase can benefit a significant portion of these patients.

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K. Shamaei and A.M. Dollar are with the School of Engineering and Applied Science, Department of Mechanical Engineering and Materials Science, Yale University, New Haven, CT 06511 USA (e-mail: {kamran.shamaei, aaron.dollar}@yale.edu).

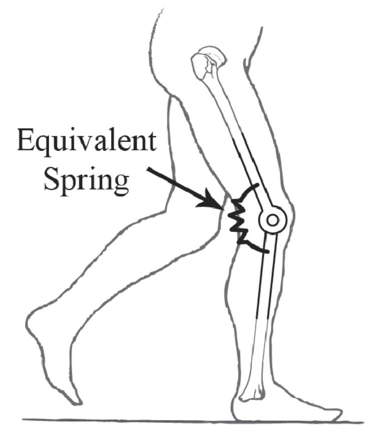

Figure 1. The knee behaves like a linear torsional spring in the weight acceptance stage of the stance phase.

To date, passive orthoses have been the primary assistive device for these types of treatments (as described in [9]). However, a number of active orthoses have recently been commercialized to treat these conditions, and more complex exoskeleton-like devices have been investigated in research ([10]-[12]). Among commercial active orthoses, stance control orthoses have been designed for patients with paresis and paralysis in the lower limb muscles. These orthoses provide the knee with necessary stability in the stance phase and allow free movement in the swing phase [13].

Stance control orthoses provide increased walking speed and cadence for novice users ([14] and [15]), and increased knee range of motion, stride, step lengths, and user satisfaction for both experienced and novice users ([13]-[15]). Moreover, users have demonstrated reduced energy expenditure and gait asymmetry [13]. However, since the knee is rigidly locked in the stance phase, users cannot demonstrate a natural gait. Particularly, the gait moves toward a compass gait and the knee does not demonstrate a shock absorption mechanism, both of which require compensatory movements that drastically increase metabolic cost, user pain and discomfort. Therefore, an orthosis that mimics the natural behavior of the support limb during stance can be highly beneficial.

As shown in section III, the moment-angle relationship of the knee joint is nearly linear during the stance phase, implying that the knee functions like a spring in this phase. Researchers postulate that external impedance parallel to the

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TABLE I
DEFINITION OF THE SYMBOLS

K_f	Characteristic stiffness of the knee in flexion mode
K_e	Characteristic stiffness of the knee in extension mode
$\bar{K}$	Characteristic stiffness of the weight acceptance stage
θ_f	Rotation in flexion mode (*deg.*)
θ_e	Rotation in extension mode (*deg.*)
R_f^2	Coefficient of determination of flexion mode
R_e^2	Coefficient of determination of flexion mode
M_{max}	Maximum moment
θ_f^0	Angle of zero moment in flexion mode (*deg.*)
θ_e^0	Angle of zero moment in extension mode (*deg.*)
θ^0	Angle of zero moment in weight acceptance stage (*deg.*)
E	Support work
$\bar{W}$	Mean body weight (*kg*)
W_b	Load weight (*kg*)
$\bar{V}$	Mean speed/cadence
$\overline{Age}$	Mean age
$\bar{H}$	Mean height (*m*)
$Control$	Control parameter in the original text
n	Number of the subjects

knee joint can decrease the impedance of the knee and therefore assist it [16]. This adaptability of the knee suggests that it might be feasible to introduce positive impedance (e.g. an external spring) during the stance phase and in parallel to the knee joint to substitute part of the function of the knee and assist with muscular weakness.

This paper focuses on the mechanical parameters that describe the behavior of the knee in the stance phase. We first present knee mechanics data from the literature to establish the spring-type behavior of the knee during the stance phase illustrated in Fig. 1. Next, we show how the mechanical parameters vary with respect to gait speed and total weight (including loads carried in a backpack). Finally, we discuss the design of orthoses and exoskeletons, employing the findings of this report.

II. METHODS

A. Weight Acceptance during Stance

The gait cycle can be divided into a stance phase and a swing phase, as shown in Fig. 2. The stance phase can be further divided into two stages including a weight acceptance stage (consisting of the initial contact, loading response, and mid-stance phases [5]) and a stance termination stage (consisting of the terminal stance and pre-swing phases [5]). The knee undergoes a resistive flexion and a propulsive extension mode during the weight acceptance stage. Exhibiting a shock damping mechanism ([17] and [18]), the knee applies a large moment in the weight acceptance stage [7]. Accordingly, the knee is highly prone to collapse at this stage without proper action of the musculature system or external assistance. Contrary to the stance leg, the swing leg approximately undergoes a ballistic movement [19] that does not demand a significant external effort.

This work focuses on the mechanical behavior of the knee in the weight acceptance stage of the stance phase of the gait. In particular, we separately evaluate the behavior of the knee in the flexion and extension modes of the weight acceptance stage.

B. Data and Terminology

The time profiles of the knee moment and angle are obtained from the open literature including [20]-[29]. The knee moment and angle are plotted against each other, an example of which is shown in Fig. 3 (data from [29]), with the instant of the onset and end of the flexion and extension modes displayed. The letters in Fig. 3 correspond to those displayed in Fig. 2. There are several parameters listed in Table I that are extensively used in this report. Particularly, the amount of rotation (θ_f and θ_e) in each mode can be obtained by subtracting the initial angle from the final angle of that particular mode. The intersections of the moment-angle traces with the horizontal axis are defined as the angle of zero moment or equilibrium angle (θ_f^0 and θ_e^0) for each mode. The angle of zero moment or equilibrium of the overall weight acceptance stage (θ^0) is defined as the mean of θ_{df}^0 and θ_{pf}^0. Moreover, the knee applies its maximum moment (M_{max}) at the instant of the onset of the extension mode.

The characteristic stiffness of the knee in flexion (K_f) and extension (K_e) modes are defined as the slopes of the corresponding regression lines shown in Fig. 3. The characteristic stiffness of the whole weight acceptance stage ($\bar{K}$) is defined as the mean of the characteristic stiffness of the flexion and extension modes. The supportive work (E) is the amount of work that the knee provides in the weight acceptance stage of one cycle of the gait in order to prevent the collapse.

Table II includes the demographic and gait properties of the individual subjects and sets of subjects for which the moment-angle relationships are extracted (with terms defined in Table I). The units of the parameters of the original texts are left intact. The reason is twofold: first, using the same units allows for comparison and avoids inconsistency. Second, in order to use identical units, the physical characteristics as well as the moment and angle time profiles of the individual subjects are required, which is usually not the case. Instead, the mean and standard deviation of the parameters and physical properties of the groups (rather than individuals) are often provided.

III. RESULTS

In this section we evaluate the variability of the mechanical parameters of the knee with respect to gait speed and weight. Section III.A. shows the knee moment/angle behavior for normal level-ground walking as collected from a number of sources in the literature. Section III.B. evaluates the effect of walking gait speed on the mechanical parameters of the knee and primarily employs the moment and angle data (rows 5 to 9 of Table II) presented by Holden et al [21]. In a similar fashion, section III.C. investigates the variability of the mechanical parameters of the knee with

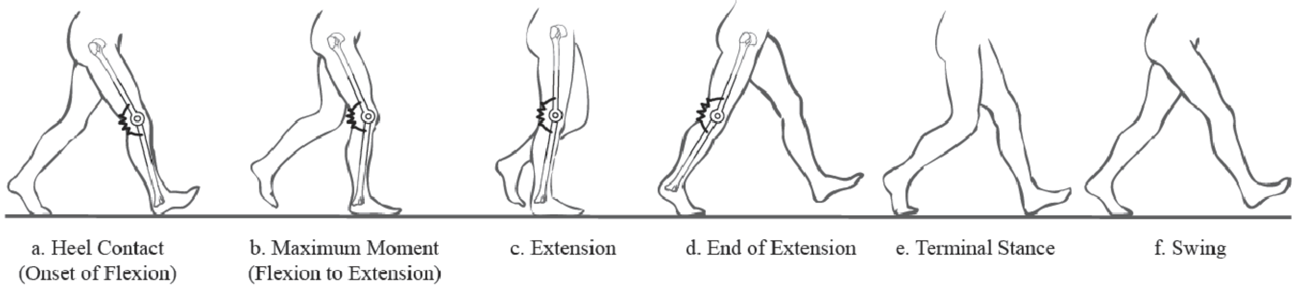

Figure 2. The knee flexes and extends under the weight of the body and then it prepares for the swing phase.

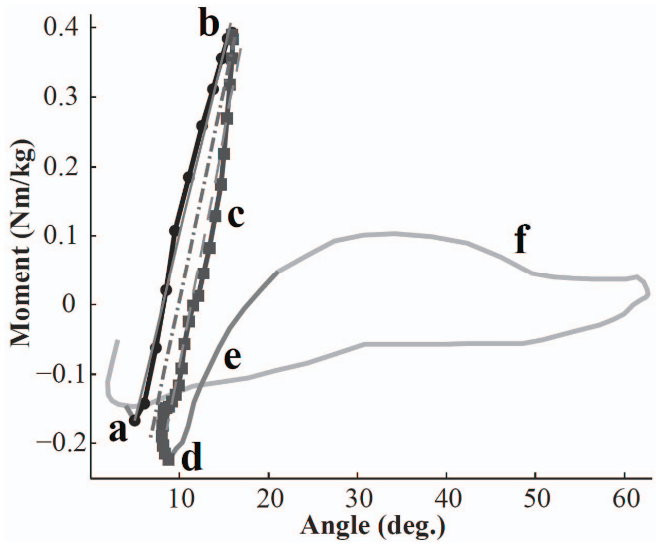

Figure 3. The knee moment-angle graph of one gait cycle for a set of subjects walking with slow cadences (85steps/min), [29]

respect to the load weight (as an indicator of the total weight) and primarily employs the moment and angle data (rows 1 to 4 of Table II) presented by Harman et al [20].

A. Knee as a Spring during Weight Acceptance Stage

Linear fits, similar to that shown in Fig. 3, are applied to the flexion and extension modes of the stance. High coefficients of determination (R^2) values of the linear regressions shown in Table II, both in flexion and extension modes, and low supportive work (E) indicate that a linear fit (spring-like behavior displayed in Fig. 1) explains most of the variability in the data. In fact, 22 of 24 subject groups have R^2 greater than *90%* in flexion mode, as shown bolded in Table II. Furthermore, 19 of 24 subjects have R^2 greater than *90%* in the extension mode, similarly, shown bolded in Table II. This finding coincides with the results of [32]. Thus, the knee can be adequately characterized, from a mechanical point of view, by a torsional spring during the weight acceptance stage of the stance phase.

Consequently, a device (including orthoses, exoskeletons, prostheses, and biped robots) can mimic the behavior of the human knee by utilizing a spring with a suitable stiffness in the stance phase and disconnecting the spring in the swing phase. Since realization of a variable stiffness mechanism is difficult to achieve, the overall characteristic stiffness ($\overline{K}$) of the weight acceptance stage is proposed as the spring constant of the envisioned device. As such, the knee is approximately modeled by a single torsional spring with stiffness equal to the mean of the stiffness of the flexion and extension modes. The characteristic stiffness of the knee in the flexion and extension modes, and in the weight acceptance stage of the gait is recorded for 24 subject groups in Table II. The characteristic stiffness of the knee would have a value from *50* to *745Nm/rad*.

If the knee is modeled by a torsional spring during the weight acceptance stage, the angle of zero moment would be the angle at which the spring is at its rest configuration (the angle at which the spring is engaged).Table II shows that the angle of zero moment range from *6* to *32deg.* The amount of rotation in flexion and extension modes defines the magnitude of deflection of the spring in the corresponding mode. Table II shows that the knee rotates from *2* to *23deg.* in the weight acceptance stage. The spring, then, applies its maximum moment when the deflection is maximal (the end of the flexion mode).

B. Knee Mechanics with respect to Gait Speed

In this section, the effect of gait speed on the mechanical parameters of the knee is investigated. These parameters include the characteristic stiffness, amount of rotation, angle of zero moment, and maximum moment of the knee.

The characteristic stiffness of the knee is plotted with respect to gait speed in Fig. 4. The data are primarily extracted from [21]. The subjects' gait speeds are adjusted to be a function of the height of the subject (*0.785Stature/sec.)* as an indicator of the natural gait speed. The characteristic stiffness of the flexion and extension modes and the overall characteristic stiffness of the weight acceptance stage are shown with black solid, dark gray dashed, and light gray dashed lines, respectively. The speed of *25%Ref.*, showing inconsistency with other speeds, is removed from the original data (Holden et al highlight this inconsistency [21]).

Linear regression shows that the characteristic stiffness of the knee changes linearly as the gait speed changes ($R^2 = 98.7, P = 0.007$ for both flexion and extension modes). It is noticeable that the changes in stiffness are in two different fashions for the flexion mode and the extension mode; particularly, the characteristic stiffness of the knee in flexion mode linearly increases as the gait speed increases; whereas the characteristic stiffness of the knee in the extension mode decreases as the gait increases. The stiffness of these two modes is predicted to be identical for a gait speed around *90%Ref.* (the intersection of the regression lines of the flexion and extension modes shown in Fig. 4) which is close

TABLE II
PHYSICAL PROPERTIES OF THE SUBJECT SETS AND MECHANICAL PARAMETERS OF THE KNEE

	Properties of Subject Groups							Mechanical Parameters												Ref.
	$\bar{W}$	$\bar{V}$‡‡	$\overline{Age}$	$\bar{H}$	Sex	n	$Control$	K_f	K_e	$\bar{K}$	θ_f	θ_e	R_f^2	R_e^2	θ_f^0	θ_e^0	θ^0	M_{max}	E	
1†	76.8	99.8C	30.3	1.81	M	16	6kg	**351**	**222**	286.5	12	17	**98**	**97**	15	11	13	53.4	-4.5	[20]
2†		100.6C					20kg	**406**	**260**	333	12	19	**98**	**98**	16	12	14	61	-4.5	
3†		100.0C					33kg	**532**	**286**	409	13	22	**98**	**98**	19	13	16	92.0	-8.4	
4†		102.4C					47kg	**553**	**322**	437.5	13	23	**96**	**98**	21	15	18	105.1	-10.5	
5•	69.7	25NS	28.8	1.72	M/F	18	25%	**11.1**	**16.7**	19.4	2	2	**90**	**80**	6	6	6	0.35	0.0	[21]
6•		50NS					50%	**14.8**	**23.8**	19.3	5	3	**95**	**94**	7	7	7	1.01	0.0	
7•		75NS					75%	**17.8**	**22.7**	20.3	10	8	**99**	**98**	8	9	8	2.18	-0.1	
8•		100NS					100%	**23.0**	**20.8**	21.9	14	12	**99**	**99**	11	11	11	3.53	0.0	
9•		125NS					125%	**28.0**	**19.1**	23.5	16	18	**99**	**94**	13	13	13	5.34	0.2	
10‡	*	*	64.4	*	M/F	37	Normal	**3.02**	**3.02**	3.02	15	13	**99**	**99**	11	11	11	0.47	0.0	[22]
11‡	70.8	120C	27.9	1.71	M/F	12	Level	**14.89**	**6.25**	5.98	9	11	**80**	**98**	19	11	15	1.16	-0.1	[23]
12‡							Downhill	**6.21**	**9.55**	7.88	10	30	**96**	**91**	*	*	*	2.58	-0.3	
13‡	64.7	1.34S	28.3	1.73	M/F	20	Healthy	**4.90**	**3.7**	4.3	23	23	**90**	**90**	14	18	16	1.86	0.3	[24]
14†	82	*	28	1.76	M	26	EE	**253**	**100**	176.5	*	*	**90**	**86**	*	*	*	*	*	[25]
15†	67.2	96C	23	1.75	M	1	A	**50**	**146**	98	15	16	**75**	**91**	26	38	32	58	-8.5	[26]
16†	70	104C	26	1.8	M	1	B	**181**	**264**	222.5	23	17	**99**	**94**	25	32	28	45	-1.4	
17†	84.3	109C	29	1.81	M	1	C	**745**	**259**	502	10	12	**85**	**89**	14	*	*	77.88	0.3	
18†	79	1.21S	21	1.82	M	1	A	**157**	**223**	190	12	7	**96**	**88**	18	20	19	13.89	-1.4	[27]
19†	63	1.18S	26	1.76	M	1	B	**147**	**250**	198.5	10	9	**98**	**98**	12	16	14	18.92	0.1	
20†	56	1.40S	21	1.62	F	1	C	**122**	**122**	122	12	12	**97**	**99**	24	32	28	34.49	0.4	
21‡	71.5	85C	22.2	1.77	*	14	Slow	**3.05**	**3.79**	3.42	11	8	**98**	**98**	8	12	10	0.39	-2.8	[29]
22‡	69.1	105C	25.6	1.75	*	16	Normal	**2.92**	**2.92**	2.92	16	14	**99**	**97**	11	13	12	0.54	0.0	
23‡	71.5	125C	22.2	1.77	*	14	Fast	**5.12**	**3.39**	4.25	18	18	**98**	**88**	7.8	13	10	1.11	1.0	
24‡	76.4	111.8C	68.9	1.71	*	18	Elderly	**3.62**	**3.27**	3.44	13	12	**99**	**95**	12	15	13	0.54	0.0	[28]

†: Stiffness *(N.m/rad.)*, Moment (*N.m*), and Work (*J*)
•: Stiffness (*%BW.Ht./rad.*) and Moment (*%BW.Ht.*), work (*%BW.Ht.*)
*: Not specified
‡: Stiffness (*N.m/kg.rad.*), Moment (*N.m/kg*), and Work (*J/kg.*)
‡‡: Cadence (*steps/min*), Speed (*m/sec.*), and Normalized Speed (*%0.785Stature/sec.*)
Notice that the subject groups are divided by the lines.

to the natural walking speed. In other words, ***the characteristic stiffnesses of the knee in flexion and extension modes are nearly identical at preferred walking speed and differ as the gait speed deviates from the natural value.***

The variations of the characteristic stiffness of the knee has two implications for the design of stance control orthoses. First, the knee does not exhibit a single characteristic stiffness for all walking speeds. As shown earlier, the characteristic stiffness of the knee changes as the gait speed varies. In an ideal case, the design of a parallel assistive device for a wide range of walking speeds would implement a variable stiffness mechanism to allow the characteristic stiffness of the knee in the flexion mode to increase and in the extension mode to decrease, with some added damping, as the walking speed increases. However, creating a truly variable stiffness mechanism is challenging in practice. Alternatively, the design of assistive devices and bipedal robots might implement a spring with the stiffness that is some weighted average of the expected range of gait speeds.

In addition to the stiffness, the overall knee flexion and extension, zero moment angle, and maximum moment also change with gait speed. The amount of rotation of the knee linearly increases ($R^2 = 94.6, P = 0.027$ for the flexion mode and $R^2 = 99.9, P = 0.001$ for the extension mode) as gait speed increases, as shown in Fig. 5. Therefore, the design of assistive devices should allow for greater rotation when the user walks faster.

The angle of zero moment increases when the subject walks faster, as shown in Fig. 6. This parameter has a linear relationship with gait speed ($R^2 = 99.7, P = 0.001$ for the flexion mode and $R^2 = 99.8, P = 0.001$ for the extension mode). The angle of zero moment represents the angle of engagement of the spring of the assistive device. Thus, the engagement of the spring of the assistive device would be delayed when the subject walks faster.

The knee also generates a larger moment as the subject walks faster. Fig. 7 shows that the maximum moment of the knee linearly increases as the gait speed increases ($R^2 = 99.0, P = 0.005$). Moreover, the value of E does not show any notable change with respect to the gait speed, hence the knee exhibits a nearly passive behavior for different gait speeds.

C. Knee Mechanics with respect to Weight

This section evaluates the variation of the mechanical parameters of the knee (the same parameters presented in section II.B.) with respect to the load weight carried by the subject. The data are primarily collected from [20] wherein the variations of the kinetics and kinematics of the gait against the load weight carried by the soldiers are presented. Harman et al. present these data for four load weights including: *6Kg*, *20Kg*, *33Kg* and *47Kg* [20]. Fig. 8 shows a linear relationship between the characteristic stiffness of the knee and the weight, both in the flexion and extension modes ($R^2 = 92.9, P = 0.036$ for the flexion mode and $R^2 = 99.7, P = 0.001$ for the extension mode). The characteristic stiffness of the flexion and extension modes and the overall characteristic stiffness of the weight acceptance stage are shown with black solid, dark gray dashed, and light gray dashed lines, respectively. The value

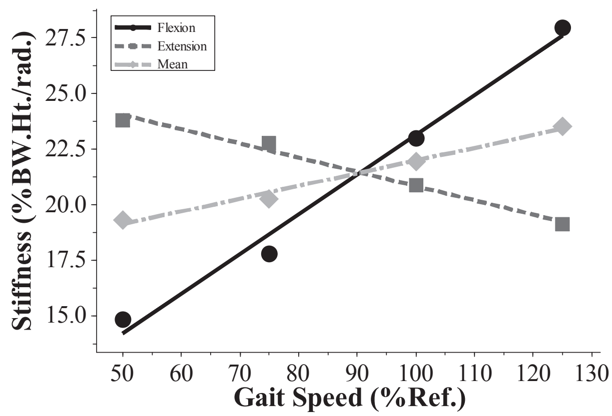

Figure 4. The characteristic stiffness of the knee in the flexion ($K_f = 5.276 + 0.178\bar{V}, R^2 = 98.7\%, p = \{0.059, 0.007\}$) and extension ($K_e = 27.230 - 0.064\bar{V}, R^2 = 98.7\%, p = \{0.000, 0.007\}$) modes, and in the weight acceptance stage ($K = 16.250 + 0.057\bar{V}$) of the gait plotted against the gait speed, (Ref. = 0.785 Stature/sec.).

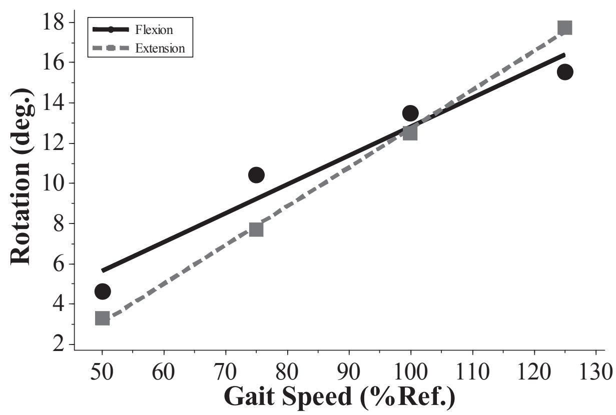

Figure 5. The amount of flexion ($\theta_f = -1.51 + 0.14\bar{V}, R^2 = 94.6\%, p = \{0.567, 0.027\}$) and extension ($\theta_e = -6.54 + 0.19\bar{V}, R^2 = 99.9\%, p = \{0.005, 0.001\}$) of the knee vs. the gait speed, (Ref. = 0.785 Stature/sec.).

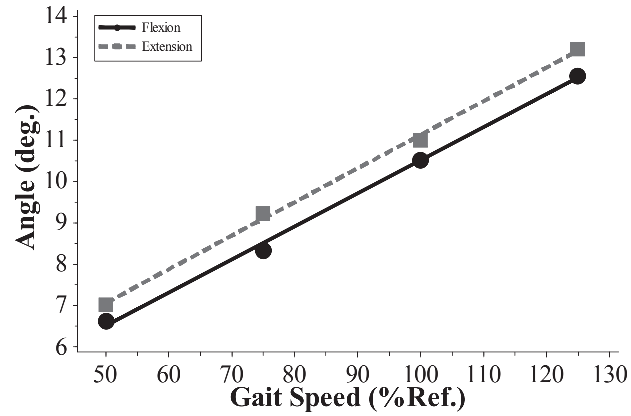

Figure 6. The angle of zero moment of flexion ($\theta_f^0 = 2.50 + 0.08\bar{V}, R^2 = 99.7\%, p = \{0.011, 0.001\}$) and extension ($\theta_e^0 = 2.99 + 0.08\bar{V}, R^2 = 99.8\%, p = \{0.005, 0.001\}$) modes plotted against the gait speed, (Ref. = 0.785 Stature/sec.).

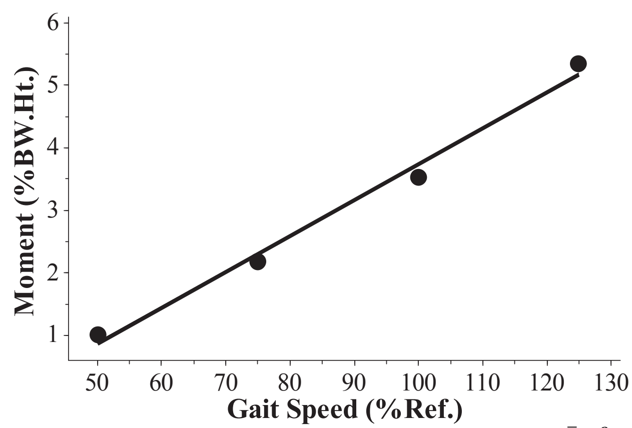

Figure 7. The maximum moment ($M_{max} = -2.004 - 0.057\bar{V}, R^2 = 99.0\%, p = \{0.035, 0.005\}$) of the knee in the weight acceptance stage vs. the gait speed, (Ref. = 0.785 Stature/sec.).

of E shows an increase as the weight increases, however, no significant linear trend of dependence could be noticed ($R^2 = 89.1, p = 0.056$) between E and W_b.

We can further plot the normalized characteristic stiffness of the knee (stiffness over the total weight) versus the load weight, as shown in Fig. 9. It can be seen that the normalized characteristic stiffness remains mostly constant for different load weights. This has several implications. First, the knee joint exhibits adaptability over this range of the total weight. In other words, as the weight increases, the knee recruits more and more muscle fibers such that the normalized characteristic stiffness of the knee remains constant. We can postulate that the knee would exhibit similar adaptability upon the presence of a parallel assistive device. However, this must be experimentally evaluated. It has been experimentally shown that, in hopping, the knee stiffness changes in presence of a serial stiffness such that the overall stiffness of the system (the knee and the surface) remains constant [30]. Along this line, others postulate that the knee, in presence of a parallel stiffness, modifies its characteristic stiffness [16].

Second, the total weight of the user should be considered in the design of a parallel assistive device. Indeed, the stiffness of the parallel assistive device should be modified as the load or pilot weight changes.

Third, the design of an ideal assistive device would employ a variable stiffness mechanism that allows for increase in the characteristic stiffness of the knee if the weight increases. However, as discussed earlier, implementation of an ideal variable stiffness mechanism is not readily achieved in practice.

The amount of flexion and extension is plotted with respect to different load weights in Fig. 10. It can be seen that the amount of flexion remains constant as the load weight increases. However, the amount of extension linearly increases as the load weight increases, ($R^2 = 96.5, P = 0.018$). The maximum rotation of the joint of the parallel assistive orthosis might be chosen as the maximum amount of flexion or extension.

Fig. 11 shows a linear relationship between the angle of zero moment and the load weight ($R^2 = 97.4, P = 0.013$ for the flexion mode and $R^2 = 96.0, P = 0.020$ for the extension mode). This variability implies that the onset of the stance control by an orthosis (which preferably would begin at the instant of zero moment) should be delayed as the load weight increases.

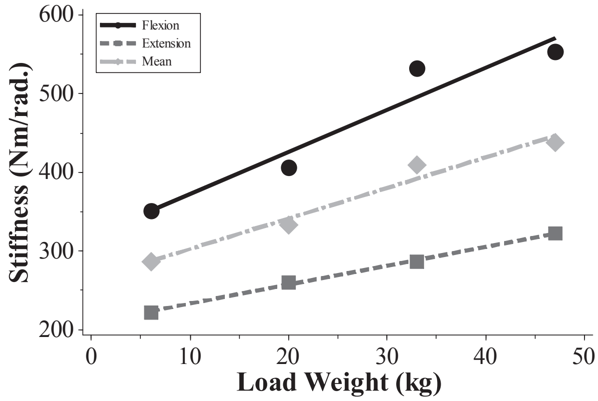

Figure 8. The characteristic stiffness of the knee in flexion ($K_f = 318.4 + 5.362W_b, R^2 = 92.9\%, p = \{0.010, 0.036\}$) and extension ($K_e = 208.9 + 2.399W_b, R^2 = 99.7\%, p = \{0.000, 0.001\}$) modes, and the weight acceptance stage ($\overline{K} = 263.7 + 3.881W_b$) of the gait plotted against the load weight.

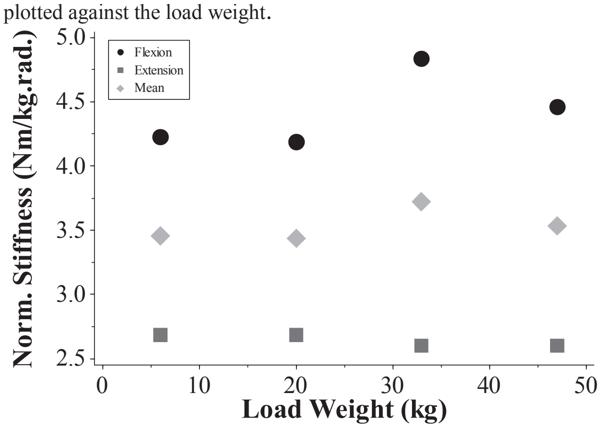

Figure 9. The normalized characteristic stiffness of the knee plotted with respect to the load weight.

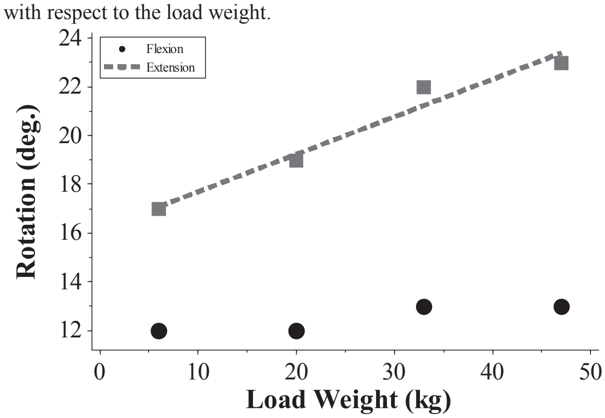

Figure 10. The amount of flexion ($\overline{\theta}_f = 12.5$) and extension ($\theta_e = 16.17 + 0.15W_b, R^2 = 96.5\%, p = \{0.002, 0.018\}$) of the knee vs. the load weight.

The variation of the maximum moment of the knee in the weight acceptance stage is plotted with respect to the load weight in Fig. 12. The plot shows a linear increase in the maximum moment as the weight increases ($R^2 = 94.3, P = 0.029$).

IV. Conclusions and Future Work

The focus of this work is on the performance of the knee in the weight acceptance stage of the stance phase of the gait and establishes the spring-type behavior of the knee during this critical stage. The mechanical parameters that accurately define the moment/angle behavior of the knee in this stage

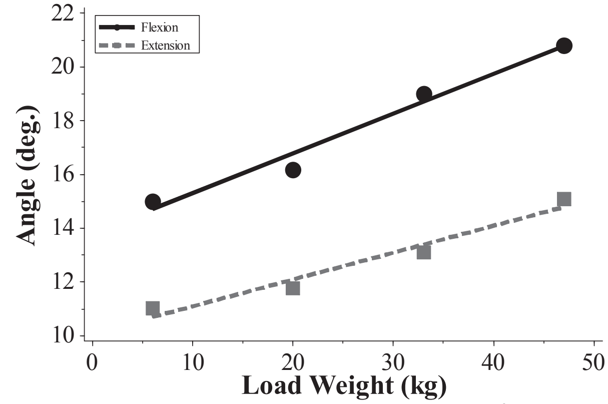

Figure 11. The angle of zero moment of flexion ($\theta_f^0 = 13.81 + 0.15W_b, R^2 = 97.4\%, p = \{0.001, 0.013\}$) and extension ($\theta_e^0 = 10.08 + 0.10W_b, R^2 = 96.0\%, p = \{0.002, 0.020\}$) modes plotted against the load weight.

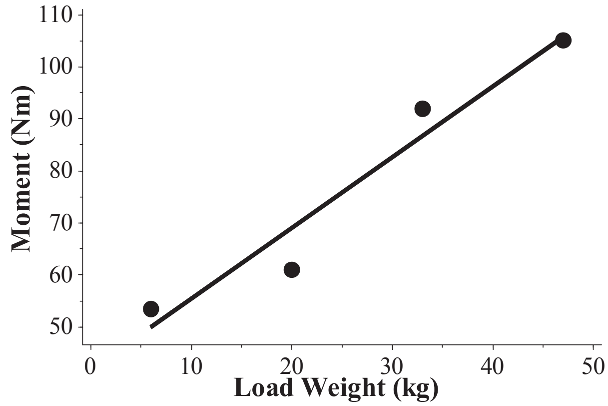

Figure 12. The maximum moment ($M_{max} = 41.740 + 1.363W_b, R^2 = 94.3\%, p = \{0.029, 0.029\}$) of the knee in the weight acceptance stage vs. load weight.

are accordingly determined. The gait speed and weight are chosen as two major parameters that affect the mechanical parameters of the knee. Finally, the changes of the mechanical parameters of the knee in the weight acceptance stage are investigated with respect to the gait speed and weight.

The current paper suggests that a parallel device (e.g. orthosis or exoskeleton) could assist the knee in stance phase if it behaves similar to a torsional spring. It is shown, however, that the stiffness, angle of engagement, and amount of rotation of the device joint should be deliberately chosen based on the gait speed and the pilot/robot weight. Ideally, a variable stiffness mechanism that adjusts its stiffness based on the gait speed and weight would be implemented to accurately mimic the behavior of the human knee joint; nevertheless, the implementation of such a device is challenging in practice.

The characteristic stiffness of the knee in the flexion mode and extension mode are predicted to be identical at the natural, preferred gait speed. This is in accordance with the results of Cavagna who shows that the rate of energy recovery is highest when the subject is walking with a natural gait speed [31]. In other words, in order for the knee to provide a high recovery rate, the knee may tend to have identical characteristic stiffness in flexion and subsequent extension modes. The characteristic stiffness of the knee at the natural gait speed is suggested to be used in the design of assistive devices.

The current work employs the kinetic and kinematic data of both the individual subjects and subject groups to establish the linear behavior of the knee in the weight acceptance stage. The open literature does not provide a complete investigation of the gait data of the individual subjects under different gait speeds and weights. However, the kinetic and kinematic gait data of the subject groups walking with different speeds and load weights are reported.

It is clear from the results presented that the mechanical parameters of the knee spring (stiffness, angle of engagement, maximum travel and moment) change significantly from one study to the next and it is not possible to correlate these to individual subject parameters using the currently available literature. Further research over a wide range of subjects in a controlled setting is required in order to predict those correlations and allow a spring to be chosen for a particular subject or range of subjects.

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