Proceedings of the 2021 Design of Medical Devices Conference
DMD2021
April 12-15, 2021, Minneapolis, MN, USA

DMD2021-1024

STUDY OF THE VARIATION IN KNEE JOINT MUSCLE FORCES AT DIFFERENT WALKING SPEEDS AND EFFECTIVENESS OF USING KNEE BRACES

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ABSTRACT

A knee assistive device can serve a large number of people to overcome muscle weakness due to aging and perform normal functional activities of the knee joint. The objective of this paper is to study the effectiveness of a knee brace to aid with normal functions and reduce muscle forces at different walking speeds. The study uses two major muscle groups, rectus femoris (RF) and biceps femoris (BF), for the muscle force study. The Electromyography (EMG) signal was recorded at 3 different walking speeds (slow, normal, fast) under both brace-assisted and normal walking conditions. EMG signals were processed and converted into muscle activation signals and finally used for muscle force calculation. The amount of assistance to each muscle group at different walking speeds was derived and analyzed to see the effectiveness of a knee brace. The knee brace was seen to have provided additional support for motion generation. However, the amount of support is found to be speed-dependent and has a different effect on different muscle groups. At slower speed, the BF muscle was seen to work against the brace, while it was not the case for the RF muscle group.

Keywords: Muscle forces, Muscle activation, EMG, Knee brace effectiveness

1. INTRODUCTION

The knee joint plays a very important role in living an active life. Quality of the life decreases with age as people lose their strength to perform their basic daily activities. The reduced functionality and strength of the knee joint are associated with difficulties in sitting, standing, and walking [1]. The decrease in muscle power and bone density, drying out and stiffening of cartilage, inflammations of the tendons, and some biological changes like less synovial fluid production are some other reasons for decreased knee joint performance [2, 3]. As a result, older people need additional support which can be provided through assistive devices. The amount of support needed for the knee joint varies from person to person depending on the muscular dynamics, joint conditions, and the kinematic and dynamic characteristics of the knee, which change significantly with age. These changes directly alter the joint load required for performing any specific task. Studies have shown that the joint load for any specific task depends on the level of activity. For example, the peak tibial force during treadmill walking was found to be around 2.1, 2.8, and 4.2 times the body weight (BW) for the speed of 1 to 3 miles/hrs, 4 miles/hrs, and 5 miles/hrs of walking, respectively [4].

The joint loads during the gait cycle are produced primarily by muscle forces [5]. Changing the muscle coordination pattern during activity through assistive devices can reduce the joint reaction forces substantially [6]. These assistive devices provide the additional force to support the natural human knee joint motion. It also helps to enable the restricted knee joint functional movements which were otherwise compromised due to aging or injuries[7]. The amount of assistive energy exerted by an assistive device depends on the compliance of the device to the human anatomical joint and musculoskeletal structure. The efficiency of these devices is measured as power support for performing a specific activity [7, 8]. This power can be used for reducing both muscle forces and joint reaction forces, altering the musculoskeletal coordination pattern for motion generation [6]. Changes in musculoskeletal coordination patterns influence muscle force-length-velocity behavior. However, this variation in joint forces is not directly related to the muscle force [9] and needs to be studied separately for better analysis. There have been several types of research on knee assistive devices from different aspects varying from assistive strategies, power augmentation strategies, locomotion control strategies, actuation systems, and control strategies for improving the knee joint assistance effectiveness. These studies have successfully quantified the amount of assistance provided through the assistive device but are yet to address the response of specific muscle groups under the provided assistance. So, it is also necessary to study the major knee joint motions and contributing muscles for providing efficient support through assistive devices.

The knee joint has three major motions, namely, extension, flexion, and rotation. These motions are achieved mainly by three major muscle groups (popliteus, quadriceps, and hamstrings), which are also responsible for the balancing and stability of the knee joint. Quadriceps (e.g. RF and vastus muscle groups) and sartorius are primary muscles responsible for producing knee extension. Knee flexion is achieved primarily by the contraction of the hamstrings muscles (BF, semimembranosus, and semitendinosus), while the popliteus muscle synergizes the motion by unlocking the fully extended knee joint. The BF muscle also helps in achieving lateral rotation whereas medial rotation is achieved mainly by Popliteus and Sartorius muscles [10]. The understanding of the synergy of these muscles along with the individual muscle force is important for a better understanding of the knee joint motion.

The muscle forces cannot be directly measured in vivo environments. To calculate these forces, several static and dynamic calculation methods have been developed [11]. The muscle force for any specific muscle can be calculated as a function of the activation signal of the muscle during that activity [12]. The muscle activation can be obtained using Electromyography (EMG), in which an action potential of the skeletal muscle is recorded. The EMG signal is directly proportional to muscle activation. The EMG-driven model for muscle activation calculation, though complicated, has been extensively used by researchers for estimating muscle forces and moment [10, 13].

The advantage of using any knee assistive device can be known through the study of the muscle forces and knee joint reaction forces induced during the knee motion. The amount of these forces required for the knee motion decreases with external support. However, the reduction of the forces is dynamic, varying with the level of activity [14]. Also, the amount of support provided by a brace in each muscle group helps to identify whether this type of brace is useful for a specific group of users or not. As older user groups tend to walk slowly, any braces addressed to this group of people have to focus on efficiency at a slower walking speed.

The purpose of this paper is to study the force variation in two major muscle groups (RF and BF) at different levels of activity and to know the effectiveness of using a knee brace for reducing the muscle forces. In order to study the reduction in forces, the methodology followed is explained in the next section. Then, the experimental setup and the number of trials performed for collecting the required data are presented. The theoretical framework for calculating the amount of assistance through the brace is presented in a reduction in muscle forces. Furthermore, EMG data processing, muscle activation calculations, and the percentage reduction in muscle forces for both RF and BF muscle groups at three different walking speeds are explained in EMG signal analysis for muscle force study. Finally, the overall results were discussed, and the conclusion of the study was stated.

2. METHODOLOGY

The study started with an identification of the major muscle groups responsible for the extension/flexion movement of the knee joint. The RF and BF muscles were chosen for the muscle force study, and a Levitation knee brace from Spring Loaded Technology was used for assisting the knee joint. EMG signals were collected at varying walking speeds from both muscle groups at both assisted and unassisted conditions with the help of a Delsys EMG system. The walking was then characterized into three different groups, namely fast walking (velocity of walking (V_w)>1.4 m/s), normal walking (1 m/s < V_w <1.4m/s), and slow walking (V_w < 1m/s) [11]. The collected EMG signals were grouped based on walking speed and processed individually using Root Mean Square (RMS) calculation technique. The processed EMG signals were then normalized and used for muscle activation calculation. Muscle forces were then defined as functions of muscle activations. Finally, any reduction in the muscle forces due to the use of a brace was calculated as a difference in the muscle activation for both assisted and unassisted conditions.

3. EXPERIMENTAL DATA ACQUISITION

An experiment was conducted to record the EMG signals from the rectus femoris (RF) and biceps femoris (BF) muscles. A 27 years old male subject with a 1.67-meter height and 69.9-kilogram body mass was used for data collection. Two Delsys Trigno wireless EMG sensors (with EMG sensor sensing range of 11mV at 20-450 Hz bandwidth) were used for capturing the action potential of RF and BF muscles. Figure 1(a) shows the EMG sensors attached to the RF muscle and Figure 1(b) to the BF muscle while wearing the Levitation brace (a lightweight carbon fiber-based product with a spring-loaded hinge at the joint and adjustable assistance knob for changing the assistive force, produced and distributed by Spring loaded Technology). The brace adjustable knob was set at a maximum assistive force position for all data acquisition, and the proper skin preparations at the place of EMG sensor attachment were carried out to collect effective EMG data.

(a) (b) (c)

FIGURE 1: Right leg with knee brace and wireless EMG signal sensor ((a) front, (b) back, and (c) side view))

The EMG data were recorded from both muscle groups with and without the brace. At first, the EMG signals were recorded for normal walking by making the subject walk for a fixed distance of 6.7 meters at a random pace. A total of 11 trials were done to record data in this phase, giving 11 EMG signals for both RF muscle and BF muscle. Similarly, in the second phase, data were recorded while wearing a brace over 9 different trials. After collecting the data in both cases, the walking speeds were classified as slow, normal, and fast speed based on the time required to cover the same distance. After classification, five slow, two normal, and two fast walking speed signals for the assisted motion were obtained. Similarly, for unassisted motion, a total of three slow, five normal, and two fast speed data were acquired. These data were then processed, normalized, and finally used for muscle activation calculations with the help of MATLAB.

4. REDUCTION IN MUSCLE FORCES

With the help of EMG signal, the muscle force generated in any specific muscle group for any activity can be generalized as,

$$F_m = a_m * A_m * ST \tag{1}$$

where A_m is the physiological cross-sectional area of muscle, ST is specific tension (31.5 N/cm2), and a_m is muscle activation [12]. The muscle activation for any muscle group can be obtained as a function of muscle force, length, and velocity relationship. The muscle force is maximum during resting and keeps decreasing while shortened or stretched. But the muscle generates no power while at rest and as the length of the muscle changes with the motion intension, the force tends to increase for some increase in velocity and gradually decreases to zero. This change in force can be described with the help of muscle activation (a_m). It can also be calculated through the help of experimentally collected EMG signal using the relation,

$$a_m = \frac{e^{Au_j(t)} - 1}{e^A - 1} \tag{2}$$

where $u_j(t)$ is a processed EMG of muscle j at time t, and A is a nonlinear shape factor, constrained to $-3 < A < 0$, with 0 representing a linear relationship [15].

The muscle activation changes with the use of a brace. So, let us consider the new muscle activation of the selected muscle group represented as a_{mb}. This can be calculated with the experimentally collected EMG signal while using a brace. Using Equation (1), the reduced muscle force F_{mb} can be rewritten as,

$$F_{mb} = a_{mb} * A_m * ST \tag{3}$$

Thus, the reduction in muscle force is $F_m - F_{mb}$.
Also,

$$Percentage\ \ reduction\ in\ muscle\ force = \frac{F_m - F_{mb}}{F_m} \times 100\% \tag{4}$$

As A_m and ST are constant, Equation (4) can be written as,

$$Percentage\ reduction\ in\ muscle\ force = \frac{a_m - a_{mb}}{a_m} \times 100\% \tag{5}$$

5. EMG SIGNAL ANALYSIS FOR MUSCLE FORCE STUDY

The recorded EMG signals were processed, rectified to get the signal profile, and plotted for different walking speeds. After plotting the obtained data, the data were normalized for each group, and through the aid of normalized data, the muscle activations were calculated using Equation (2). After getting the muscle activations, the percentage reduction in muscle force for both muscle groups at different speeds was calculated by the use of Equation (5) in MATLAB. For comparison purposes, data peak, data average, and peak average of all data were taken and compared to the respective walking speed pair.

5.1 Rectus femoris muscle force

A study of the RF EMG signal shows that muscle activation is higher at higher walking speeds and gradually decreases with a decrease in walking speed. The comparison of the EMG signals generated during the first 4 seconds of walking at three different speeds was plotted against time and presented in Figure 2. The signal obtained during the fast walking condition (represented in red) has the highest signal strength. This is followed by a signal obtained during the normal walking condition (plotted in black) with moderate signal strength. And finally, the slow walking produced the weakest signal (shown in blue). The peak of each signal represents the maximum muscle excitation/activation during the leg push-off phase of the gait cycle. This higher muscle activation indicates the higher muscle force generated during that activity. So, this indicates that the muscle force increases with the increase in walking speed.

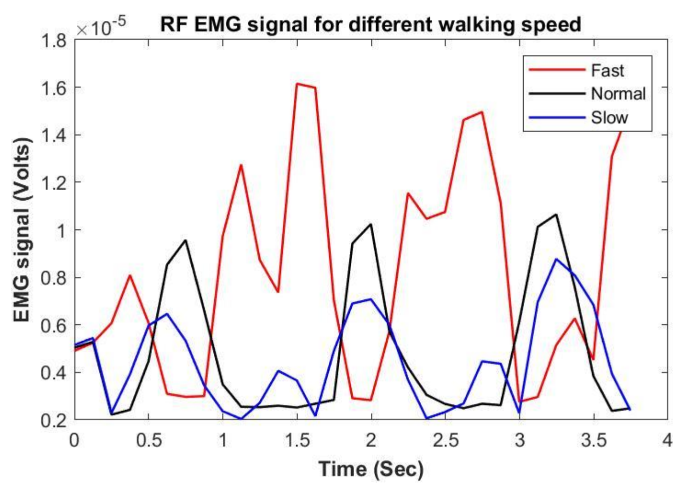

FIGURE 2: EMG signal generated in RF muscle without a brace

The same procedure was used to analyze the EMG signal while using a brace. For all walking speeds, the EMG signal was found to be of the same pattern but with slightly lower intensity compared to the respective unassisted motion. The smaller EMG signal signifies the decrease in the muscle action potential or the muscle force. This difference in EMG signal strength for each walking speed during assisted and unassisted motion can be studied in Figure 3. For fast unassisted walking conditions, the peak value of EMG is found to be around 1.6×10^{-5} volts (v), which decreases to the range of 1×10^{-5} v for assisted motion. This decrease in muscle excitation indicates a decrease in muscle force, which is also the case for normal and slow walking conditions.

After analyzing all EMG data, the RF muscle force reduction was calculated at all three speeds using peak, data average, and peak average of the data and summarized as in Table 1. To perform these calculations using the normalized activation signal, the global peak of signal for peak study, the average of the local peaks for peak average study, and the data mean for data average study were taken and compared to respective assisted/unassisted motion pair. The muscle force reduction was found to be 20.17% for slow walking during peak study, which increased with the increase in walking speed. Similarly, the force reduction was found increased for data average study and highest in peak average study for all walking speed. This shows that the amount of assistance provided is speed-dependent which decreases with a decrease in speed and vice versa. Also, the change in the muscle reduction force in different study methods indicates the dynamic nature of the muscle during activity.

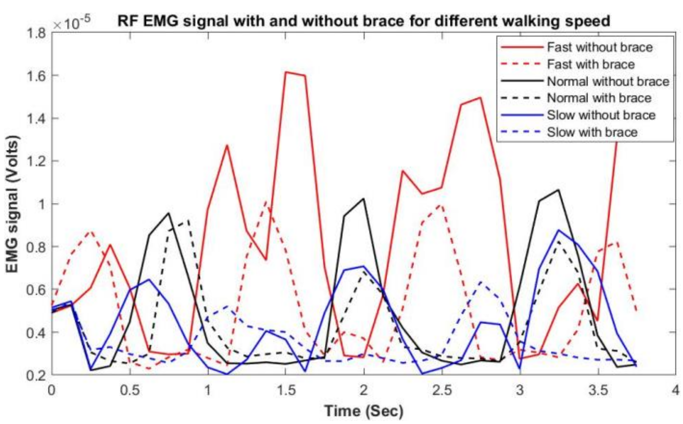

FIGURE 3: EMG signal comparison of RF muscle at fast, normal, and slow speed with and without a brace

TABLE 1: Percentage of muscle force reduction in rectus femoris muscle

Walking speed	*Peak*	*Data average*	*Peak average*
Slow	20.17	22.33	22.84
normal	23.74	25.92	27.39
fast	23.75	34.64	40.79

5.2 Biceps femoris muscle force

BF muscle behavior under the assisted condition was found different compared to the RF muscle group. Though the EMG signal intensity of RF muscle increased with the increase in the activity level (similar to RF muscle as shown in Figure 2), the assistance from the brace was found to be drastically different. In fact, at the slow walking speed, the muscle was seen to work against the brace. So, instead of aiding the functionality of the knee, it was restricting the movement making it harder for the wearer. This can be observed by looking at the EMG signal plot presented in Figure 4, where the EMG signal strength for slow walking during the assisted condition (shown in blue dot line) is higher compared to the unassisted condition (blue line).

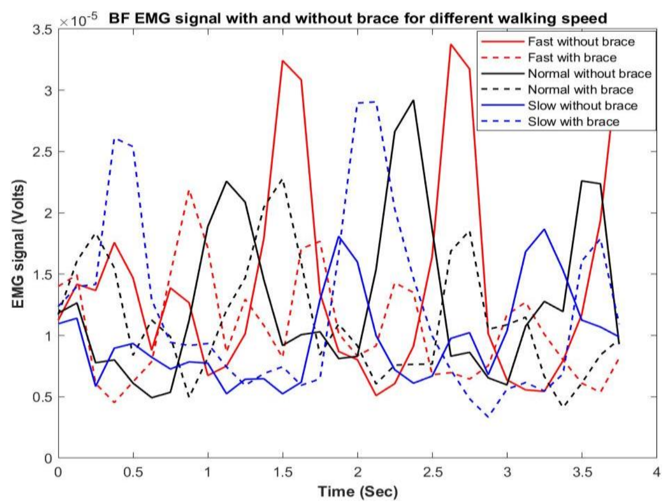

FIGURE 4: EMG signal comparison of BF muscle at fast, normal, and slow speed with and without a brace

The assistance obtained for the biceps femoris at all three speeds under different types of data study is listed in Table 2. It shows that there is an increase in BF muscle load due to the use of the assistive device, which varies from 0.98% during the data average study to 20.81% for peak average study. However, the rate of change of the assistive force with speed is found to be very high, making it effective even with a slight increase in speed. The percentage of muscle force reduction at a normal speed is found positive for all data studies, where the peak data study shows the highest assistance (33.37%) and the data average study shows the lowest assistance (4.39%). At fast speed, it shows almost 60% force reduction while comparing the peak data and almost 50% and above in other comparisons.

TABLE 2: Percentage of muscle force reduction in biceps femoris muscle

Walking speed	*Peak*	*Data average*	*Peak average*
Slow	-12.8	-0.98	-20.81
normal	33.37	4.39	29.5
fast	59.62	48.77	55.65

To study the overall advantage of the knee brace, the percentage of muscle force reduction in all walking speeds was plotted for both muscle groups and is shown in Figure 5. It shows that the percentage in muscle force reduction, or the knee brace efficiency, increases with the increase in the walking speed. While moving from slow to fast speed, the brace showed a 12-18% increase in assistive force for RF muscle during different data studies. This increase however is very high for BF varying from 49% to almost 56%.

6. DISCUSSION AND CONCLUSION

The assistive force provided by the brace to any specific muscle group was found to be dynamic and speed dependent. Generally, the amount of assistance increased with an increase in the level of activity. But the different muscle group was seen to have responded differently for some activities. In the case of RF muscle, using the brace was beneficial at all the speeds examined, and an increase in the efficiency with the increase in walking speed was gradual. But in the case of the BF muscle group, there was a vast change in the assistive behavior with the increase in speed. At fast speed, it was more supportive than in the RF muscle group, reducing the muscle force by more than half. But as the walking speed decreased, the amount of support drastically decreased and even became negative at slow speed. This indicates that, at the slow speed, instead of assisting the wearer, the muscle was working against it. This analysis can also be supported through the EMG signal analysis by looking at the plots in Figure 4, where the EMG signal strength while wearing a brace at slow speed is higher compared to normal walking.

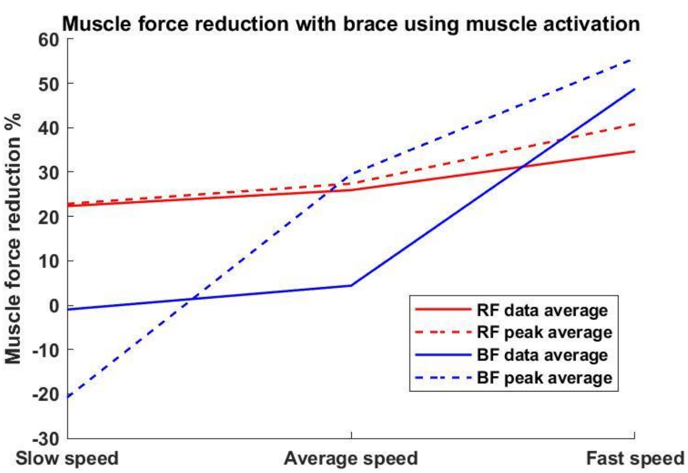

FIGURE 5: Reduction in muscle force with varying speed

In this study, the assistance provided by the knee brace to major knee muscle groups at different walking speeds was studied. The level of assistance was found to be dynamic in nature, varying largely depending on the speed and muscle group. The amount of assistance was high for faster speed and low for slower speed. Also, the assistive force at slow speed was contrasting for RF and BF but supporting for both normal and faster speed. The minimum assistive force at slow walking speed makes these mechanical braces unattractive for slow-walking elderly, creating a need for better knee assistive devices. This research can be beneficial for designing an assistive device that can redistribute the assistive force to synchronize different muscle groups and produce efficient motion. In the future, we would like to extend this research by collecting more EMG data from multiple subjects and provide more solid results to facilitate a better design of knee braces for the elderly.

ACKNOWLEDGEMENTS

The authors would like to thank Baylor University for a start-up fund provided to Dr. Ro for this research and thank Jingdong Chen and Nisha Bhattarai for their assistance during experimental data collection.

REFERENCES

[1] Kikuchi, T., Sakai, K., and Abe, I., 2016. "Bioinspired knee joint for a power-assist suit". Journal of Robotics, 2016.

[2] Kapsalyamov, A., Jamwal, P. K., Hussain, S., and Ghayesh, M. H., 2019. "State of the art lower limb robotic exoskeletons for elderly assistance". IEEE Access, 7, pp. 95075–95086.

[3] Boyer, K. A., and Andriacchi, T. P., 2016. "The nature of age-related differences in knee function during walking: implication for the development of knee osteoarthritis". PloS one, 11(12), p. e0167352.

[4] D'Lima, D. D., Fregly, B. J., Patil, S., Steklov, N., and Colwell Jr, C. W., 2012. "Knee joint forces: prediction, measurement, and significance". Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine, 226(2), pp. 95–102.

[5] Sasaki, K., and Neptune, R. R., 2010. "Individual muscle contributions to the axial knee joint contact force during normal walking". Journal of biomechanics, 43(14), pp. 2780–2784.

[6] DeMers, M. S., Pal, S., and Delp, S. L., 2014. "Changes in tibiofemoral forces due to variations in muscle activity during walking". Journal of orthopaedic research, 32(6), pp. 769–776.

[7] Lee, K.-M., and Guo, J., 2010. "Kinematic and dynamic analysis of an anatomically based knee joint". Journal of biomechanics, 43(7), pp. 1231–1236.

[8] Low, K., 2005. "Initial experiments on a leg mechanism with a flexible geared joint and footpad". Advanced Robotics, 19(4), pp. 373–399.

[9] Hoy, M. G., Zajac, F. E., and Gordon, M. E., 1990. "A musculoskeletal model of the human lower extremity: the effect of muscle, tendon, and moment arm on the moment angle relationship of musculotendon actuators at the hip, knee, and ankle". Journal of biomechanics, 23(2), pp. 157–169.

[10] Adhikari, V., Yihun, Y., and Lankarani, H. M., 2018. "Design of a novel task-based knee rehabilitation exoskeleton device". In Frontiers in Biomedical Devices, Vol. 40789, American Society of Mechanical Engineers, p. V001T03A020.

[11] Anderson, F. C., and Pandy, M. G., 2001. "Static and dynamic optimization solutions for gait are practically

equivalent". Journal of biomechanics, 34(2), pp. 153–161.

[12] Yoshioka, S., Nagano, A., Hay, D. C., and Fukashiro, S., 2012. "The minimum required muscle force for a sit-to-stand task". Journal of biomechanics, 45(4), pp. 699–705.

[13] Buchanan, T. S., Lloyd, D. G., Manal, K., Besier, T. F., et al., 2005. "Estimation of muscle forces and joint moments using a forward-inverse dynamics model". Medicine and Science in Sports and exercise, 37(11), p. 1911.

[14] Fritz, S., and Lusardi, M., 2009. "White paper: "walking speed: the sixth vital sign"". Journal of geriatric physical therapy, 32(2), pp. 2–5.

[15] Lloyd, D. G., and Besier, T. F., 2003. "An emg-driven musculoskeletal model to estimate muscle forces and knee joint moments in vivo". Journal of biomechanics, 36(6), pp. 765–776.