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Reduction of gait data variability using curve registration

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Abstract

Timing in peak gait values shifts slightly between gait trials. When averaged, the standard deviation (S.D.) in gait data may increase due to this inter-trial variability unless normalization is carried out beforehand. The objective of this study was to determine how curve registration, an alignment technique, can reduce inter-subject variability in gait data without perturbing the curve characteristics. Twenty young, healthy men participated in this study each providing a single gait trial. Gait was assessed by means of a four-camera high-speed video system synchronized to a force plate. A rigid body three-segment model was used in an inverse dynamic approach to calculate three-dimensional muscle powers at the hip, knee and ankle. Curve registration was applied to each of the 20 gait trials to align the peak powers. The mean registered peak powers increased by an average of 0.10 ± 0.13 W/kg with the highest increases in the sagittal plane at push-off. After performing curve registration, the RMS values decreased by 13.6% and the greatest reduction occurred at the hip and knee, both in the sagittal plane. No important discontinuities were reported in the first and second derivatives of the unregistered and registered curves. Curve registration did not have much effect on the harmonic content. This would be an appropriate technique for application prior to any statistical analysis using able-bodied gait patterns.

Keywords: Biomechanics; Gait pattern analysis; Continuous data; Curve registration technique

1. Introduction

Three-dimensional (3D) video-based systems combined with force plate data provide large quantities of information. These are usually time-dependent, but to simplify the analysis and facilitate data interpretation, peak values are often used [1,2] to characterize able-bodied and pathological gaits. However, due to inter-individual variability, peak values usually occur at slightly varying times within the gait cycle (GC). This explains the discrepancies between the mean peak values reported in tables and those illustrated in figures.

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As reported in the literature [3,4], gait patterns of able-bodied subjects can be assumed to be reproducible, but with some variation. When averaging trials of several subjects is done, information is lost in the means. This problem is presented in Fig. 1, where the sagittal hip power curves of seven individuals are illustrated in cascade form. The peak powers at push-off occurred within a spread of 11% ranging between the 52 and 63% marks of the gait cycle. By averaging these curves, the mean power would be 46% less (3.22 rather than 4.71 W/kg) than the average of the individual peaks. Similar observations can be made on other peak powers of this curve. Consequently, some kind of time-normalization [5,6] is necessary to facilitate comparison between individuals.

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Normalization of the curve's amplitude is routinely done by scaling the parameter values with a physical constant such as body weight, subject's height, segment circumference, etc. [7]. Time-scale normalization can also be performed by time warping techniques [8], image normalization [9] or curve registration [10,11]. Curve registration was chosen here because it can be applied to time-dependent parameters such as net joint moments, muscle powers, etc. With this technique, individual data can be adjusted to events identified in the gait cycle. This alignment technique is similar to fixing the stance phase duration of a group at its mean value [12]. To our knowledge, curve registration has never been used in gait data processing.

The objective of this study was to demonstrate how curve registration can reduce inter-subject variability developed in able-bodied subjects during gait without perturbing the muscle power curve characteristics. Among such gait parameters as ground reaction forces, muscle moments or muscle powers, the latter were used for data variability analysis because they have been shown to be one of the best indicators of a person's ability to walk [1,3].

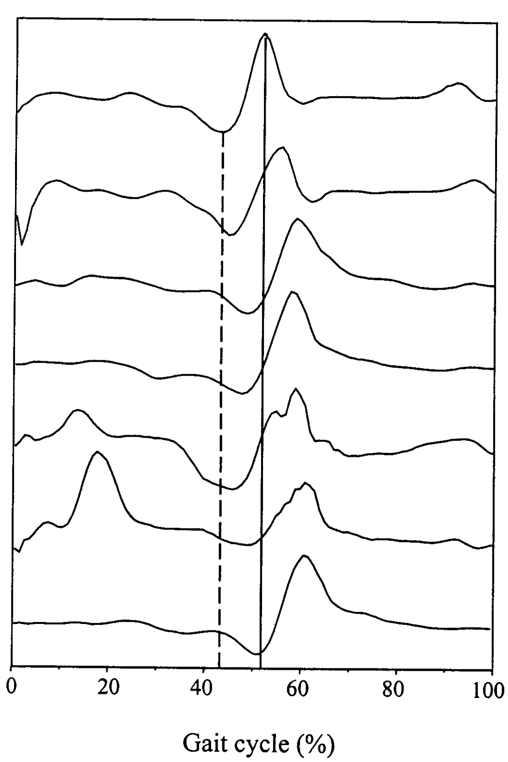

Fig. 1. Hip muscle power curves developed in the sagittal plane during a gait cycle for seven subjects. The dashed and solid lines are respectively aligned with the beginning and the peak value of the hip power generation at push-off for the subject whose curve is shown at the top.

2. Methods

Twenty young, healthy adult men with an average age of 25.3 ± 4.1 years, height of 1.77 ± 0.06 m and mass of 80.6 ± 13.8 kg participated in this study. The 3D kinematics of the right lower limb were obtained using ten reflective markers positioned on the subjects' trunk, thigh, shank and foot [2]. The gait parameters were calculated in the joint coordinate system using measurements taken between the external markers and the estimated joint center of rotation.

Data acquisition was performed with a Motion Analysis system consisting of four cameras (90 Hz) synchronized to an AMTI force plate (360 Hz). The cameras were located an average distance of 4.5 m from the walkway and along an arc of 120° to cover a complete stride. Subjects were asked to walk at a self-determined pace along a 11 m walkway and step on the force plate. A few practice trials were allowed before recording a single gait cycle.

The 3D position of the markers was obtained by the Direct Linear Transformation technique. A fourth order zero-phase lag Butterworth low-pass filter was applied to reduce the noise in the video and force plate data having respective cut-off frequencies of 6 and 30 Hz. The inverse dynamic method was employed to calculate the muscle moments at each joint and in each plane throughout the gait cycle. Instantaneous muscle powers were estimated as the product of the net muscle moments and their corresponding joint angular velocity calculated at each time instant. Joint moments and angular velocities acting in the same direction resulted in power generation, whereas power absorption was obtained when the polarities were different. Each muscle power value was normalized with respect to body mass.

For each subject, 24 muscle power bursts were identified and labeled according to Eng and Winter [4]. The first letter refers to the joint; the number indicates the sequence of the power burst and the second letter identifies the plane of motion. For example, H3S corresponds to the third power burst of the hip which occurred in the sagittal plane. The mean stance phase of the right limb of the 20 able-bodied subjects was set at 60.68% of the gait cycle (GC). Setting the stance phase at 60.68%, GC may have introduced some phase shift of its own which, if present, would still be in the data even without the registration process. However, it is important to note that the purpose here was to demonstrate how curve registration can be applied to a data set and that this process does not substantially alter the original data regardless of its initial quality.

Curve registration [10,13] consists of four steps which were performed on each of the continuous muscle power curves. For a specific joint and plane, a mean muscle power curve was calculated from the data of 20

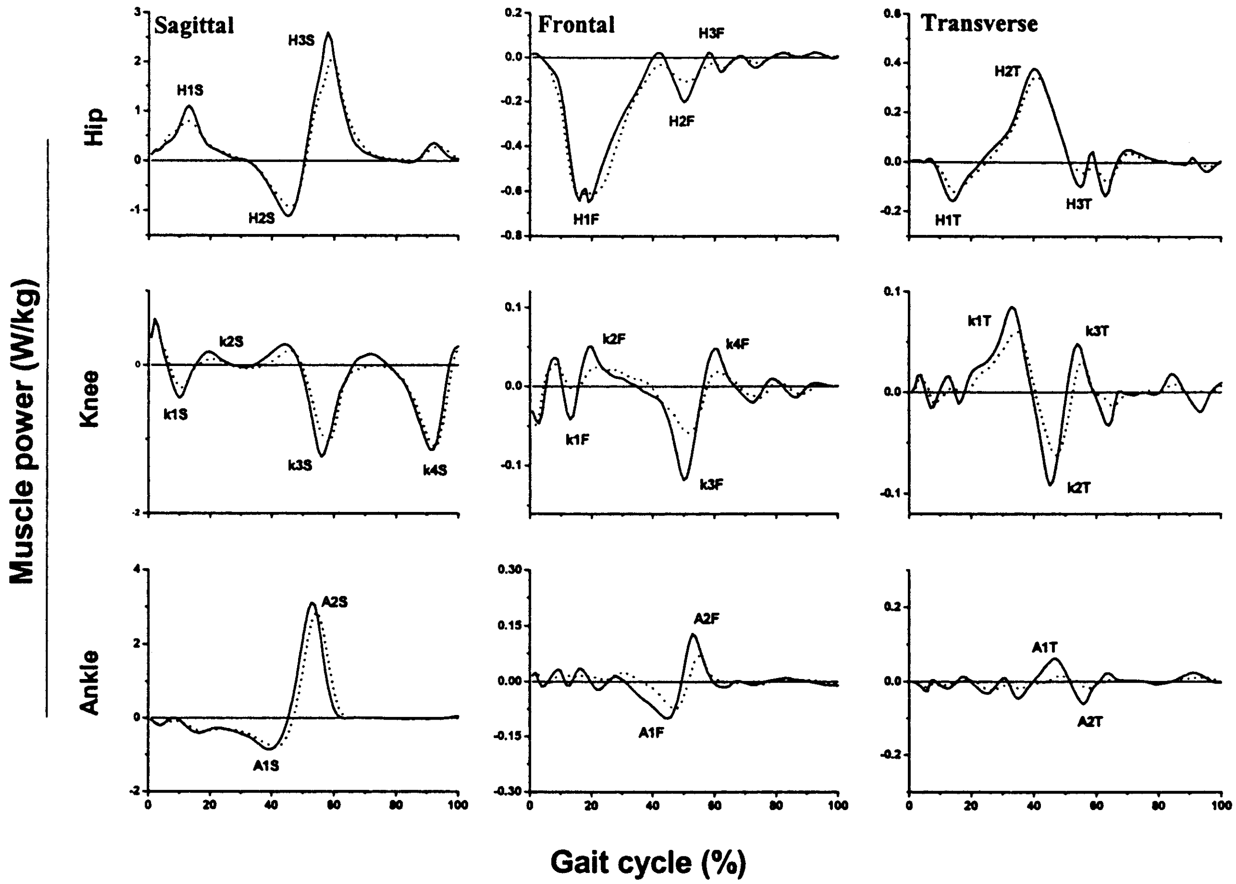

Fig. 2. Three-dimensional mean unregistered (dotted lines) and registered (solid lines) muscle power curves (W/kg) of the right hip, knee and ankle developed by the 20 subjects during a gait cycle.

able-bodied subjects. Then, on this mean unregistered muscle power curve, which formed the initial template, the timings of selected landmarks, consisting of inflection points and zero-crossings, were identified. Usually five time instants were identified per muscle power curve, at heel-strike, $t_{i1,}$ at three other time instants; t_{i2}, t_{i3}, $t_{i4,}$ according to the shape of the muscle power curve; and at the end of the gait cycle, t_{i5}. Then for each subject, the timings of the same five landmarks, j, were identified. For a given curve, the relation between the t_{ij}'s on the horizontal axis and the values on the vertical axis was obtained by linear interpolation to give a function relating the target time to the warped time. The final step consisted in interpolating the values for the entire curve at the warped times to get new values at equally spaced points, and these are the values for curve i. In other words, this new curve formed the registered curve. Finally, a mean registered curve was calculated from the 20 registered curves.

To verify to what extent curve registration modified the peak muscle powers, differences between the unregistered and registered corresponding peaks were determined using student's paired t-tests ($P < 0.05$). The effects of registration on the curve structural characteristics were also evaluated. The root mean square (RMS) difference between the unregistered and registered curves was calculated to quantify the changes in the shape of the muscle power curves. To determine the presence of discontinuities in the unregistered and registered curves, their first and second derivatives were obtained by the central difference technique. RMS differences between these unregistered and registered derivative curves were also calculated. Finally, a power spectrum analysis was performed on the unregistered and registered curves of each subject, to test, if the registration technique modified the harmonic content of the muscle power curves. Statistical differences between the unregistered and registered means and median frequencies were determined using the Student's t-test with $P < 0.05$.

3. Results

Fig. 2 presents the average unregistered and registered 3D muscle power curves developed at the hip, knee and ankle. Variations in muscle power curves were noted for all joints and all planes. S.D. curves were omitted here for clarity, though they were calculated and are reported for the peak powers in Table 1.

Seventeen of the 24 registered peak muscle powers given in Table 1 were significantly different from the unregistered ones. Sixteen were statistically higher than their corresponding unregistered values. Only the registered H3F value was reduced by 0.05 W/kg. For all subjects, the mean registered peak powers increased by 0.10 ± 0.13 W/kg. These increases were approximately equally distributed in all joints and planes. Though the mean increase was small, substantial increases were noted for, (a) the hip H3S pulling action at push-off (0.53 W/kg, 19.6%), (b) the knee K3S absorption at push-off (0.20 W/kg, 19.2%), (c) the ankle A2S push-off (0.27 W/kg, 9.5%), and (d) the hip H2S absorption during terminal stance (0.19 W/kg, 20.4%).

Table 2 provides the RMS values calculated between the mean unregistered curve and each individual unregistered curve for all joints and planes and between the mean registered curve and individual registered values, respectively. The RMS values for the unregistered curves varied between 0.003 and 0.063 W/kg while for the registered curves the RMS values were between 0.003 and 0.056 W/kg. In six out of nine curves, the RMS values of the registered curves were smaller than for the unregistered ones while for the other three, the RMS values were identical. Curve registration decreased the mean RMS values by an average of 13.6%. The greatest statistically significant reductions in the RMS values for registered means were observed for the hip (12.5%) and knee (23.0%) both in the sagittal plane. The joint which displayed the greatest mean reduction was the hip while it was in the sagittal plane that the variability decreased substantially.

To verify changes due to the registration process on the original data, the first and second derivatives were calculated. Fig. 3 shows that the amplitude of the derivatives increased with registration more in the transverse than the sagittal plane. In the frontal plane, changes were sometimes smaller than those in the other two planes. This can be verified in Table 3, where the RMS values of these derivative curves are presented. In this table, the RMS values of the registered derivative curves are larger than the unregistered ones in all planes: mean increase was 57.16 and 73.01%, respectively for the first and second derivatives. RMS values of the 1st derivative were the lowest (0.27 W/kg per step) and highest (17.12 W/kg per step) for the ankle in the transverse and sagittal plane, respectively. For the second derivative, the minimum and maximum values were again found at the ankle level in the transverse (14.98 W/kg per step^2) and sagittal (559.38 W/kg per step^2) planes.

Power spectrum analysis was also performed. Since the gait cycle had been normalized to 100%, means and medians of the spectra were expressed in number of harmonics. As shown in Table 4, means for the unregistered curves ranged between 1.30 and 2.46, while for the registered data, they were between 1.40 and 2.20.

Table 1
Mean and standard deviation (S.D.) of peak muscle powers (W/kg) for unregistered and registered data[a]

Parameters	Unregistered		Registered	
	Mean	S.D.	Mean	S.D.
H1S	0.88	0.83	1.10	1.02
H2S	−0.93*	0.79	−1.12*	0.82
H3S	2.07*	1.44	2.60*	1.26
H1F	−0.61	0.30	−0.59	0.32
H2F	−0.11*	0.07	−0.20*	0.09
H3F	−0.03*	0.02	0.02*	0.08
H1T	−0.15	0.05	−0.16	0.13
H2T	0.35*	0.27	0.38*	0.26
H3T	−0.09*	0.06	−0.16*	0.12
K1S	−0.33*	0.27	−0.45*	0.34
K2S	0.20	0.16	0.22	0.16
K3S	−1.04*	0.75	−1.24*	0.64
K4S	−1.11	0.64	−1.15	0.64
K1F	−0.02	0.09	−0.03	0.02
K2F	0.04*	0.03	0.05*	0.03
K3F	−0.05*	0.04	−0.10*	0.07
K4F	0.02*	0.02	0.04*	0.02
K1T	0.06*	0.05	0.09*	0.04
K2T	−0.06*	0.05	−0.09*	0.05
K3T	0.05*	0.04	0.06*	0.05
A1S	−0.80*	0.26	−0.87*	0.26
A2S	2.84*	0.70	3.11*	0.67
A1F	−0.06*	0.04	−0.11*	0.07
A2F	0.08	0.05	0.13	0.10

[a] Statistically significant differences are indicated by * for $P<0.05$

Table 2
RMS values (W/kg) between the mean curves and their corresponding unregistered and registered individual trials of muscle power curves of 20 able-bodied subjects[a]

Joint	Plane	RMS (W/kg)		Decrease (%)
		Unregistered	Registered	
		Mean (S.D.)	Mean (S.D.)	
Hip	Sagittal	0.063* (0.031)	0.05* (0.024)	−12.5
	Frontal	0.025 (0.013)	0.024 (0.013)	−4.2
	Transverse	0.013 (0.007)	0.013 (0.008)	0.0
Knee	Sagittal	0.038* (0.014)	0.031* (0.012)	−23.0
	Frontal	0.006 (0.007)	0.006 (0.007)	0.0
	Transverse	0.003 (0.001)	0.003 (0.001)	0.0
Ankle	Sagittal	0.036 (0.021)	0.031 (0.024)	−16.0
	Frontal	0.006* (0.003)	0.005* (0.002)	−20.0
	Transverse	0.004* (0.002)	0.003* (0.002)	−33.3

[a] Statistically significant differences are indicated by * for $P<0.05$.

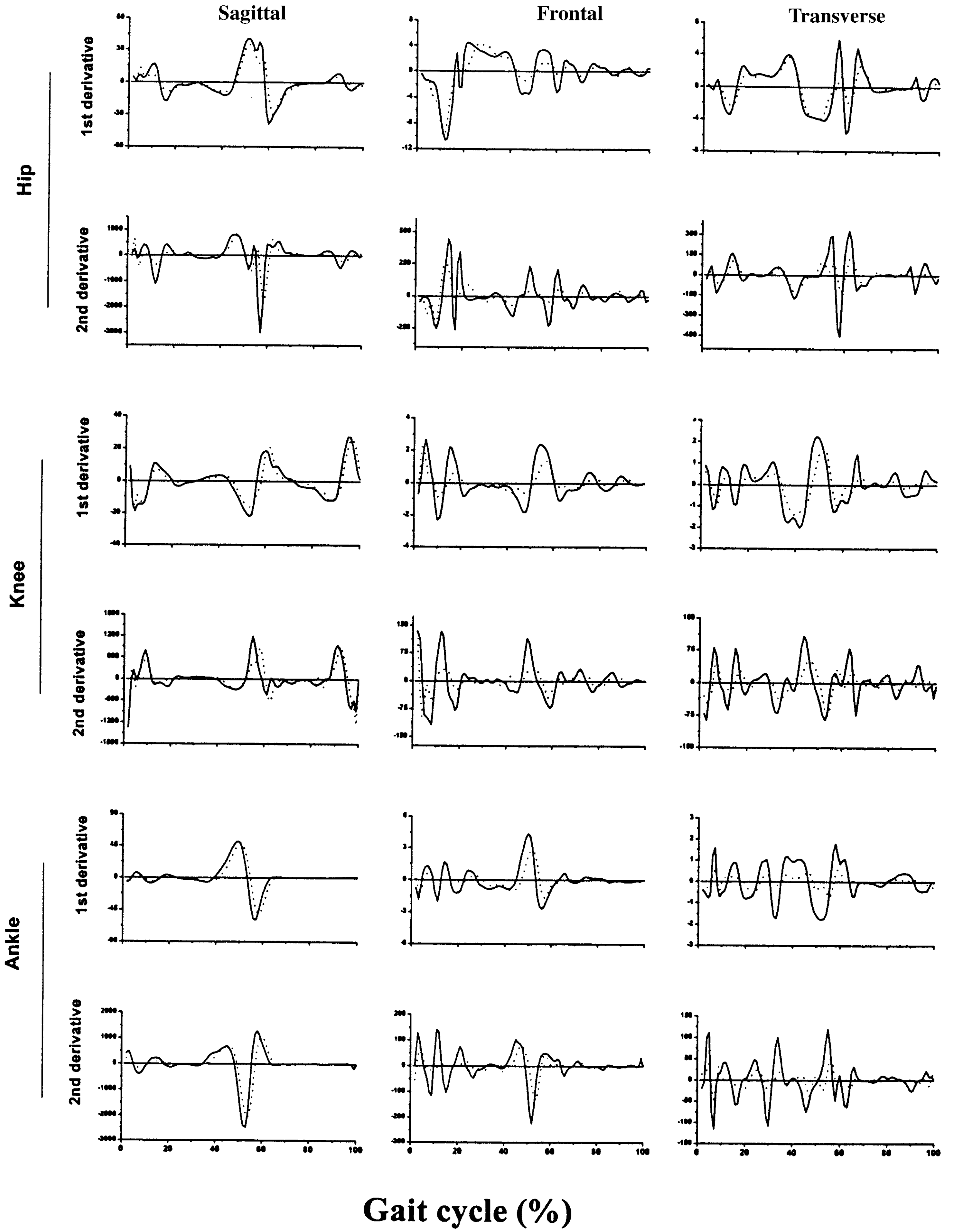

Fig. 3. First (W/kg per step) and second (W/kg per step) derivatives of unregistered (dotted lines) and registered (solid lines) of the mean muscle power curves of 20 normal subjects. Data are from the right hip, knee and ankle in the sagittal, frontal and transverse planes.

Table 3
RMS values and differences for the first and second derivatives for the unregistered and registered data

Joint	Plane	RMS					
		First derivative (W/kg per step)			Second derivative (W/kg per step)		
		Unregistered	Registered	Difference (%)	Unregistered	Registered	Difference (%)
Hip	Sagittal	11.58	14.85	28.16	374.96	520.95	38.94
	Frontal	2.50	2.99	19.22	68.33	116.28	70.17
	Transverse	1.71	2.36	38.44	52.39	109.26	108.53
Knee	Sagittal	9.06	10.34	14.09	338.14	371.03	9.73
	Frontal	0.55	1.01	82.89	23.97	45.81	91.14
	Transverse	0.59	0.91	53.74	22.88	38.99	70.40
Ankle	Sagittal	15.61	17.12	9.64	485.14	559.38	15.30
	Frontal	0.71	1.20	68.05	30.52	55.17	80.76
	Transverse	0.27	0.80	200.26	14.98	40.75	172.12

The medians varied between 2.34 and 4.78 for the unregistered data and between 2.45 and 4.45 for the registered data. No significant difference was detected in the harmonic content except for the mean and median values for the ankle in the frontal plane and for the mean of the knee in the transverse plane and the mean of the hip in the frontal plane.

4. Discussion

The objective of this study was to demonstrate how curve registration can be used to reduce inter-subject variability by normalizing time-related events occurring during the gait cycle. Curve registration requires, that the data to be treated, meet at least two criteria [10]. First, all curves should have a typical structural pattern common to all samples derived from different subjects, notwithstanding variations in both amplitude and phase between individual curves. This is an important consideration. In able-bodied gait, Vardaxis et al., [14] have shown that subjects display a high similarity between their own gait trials. Using multivariate analysis, they were able to group together a subject's individual gait trials and dissociate each subject from the others. However, this requirement may not be fulfilled when assessing other populations. For example, Watelain et al., [15] used cluster analysis to group gait trials of young able-bodied subjects and healthy elderly subjects. Discrepancies were found between the gait trials of the elderly subjects. This was attributed to the difficulty obtaining reproducible gait trials particularly for the phasic and temporal gait parameters as well as for the sagittal muscle powers. In such cases, and in certain gait pathologies such as cerebral palsy, variations in consecutive gait trials can be important and make these data unsuitable for curve registration.The authors of the present study recommend that the mean registered curve obtained from able-bodied subjects be used as a reference to compare any able-bodied gait patterns. For pathological gait where the same labeling points as in able-bodied gait patterns can be recognized, curve registration techniques may also be suitable.

The second criterion requires that landmarks chosen for curve alignment should be clearly visible and identifiable in all individual curves. These landmarks are relatively easy to identify in the sagittal plane data, where peak values are high and zero-crossing, well defined. But these events are slightly more difficult to locate in the two other planes due to the flattening of the power curves and lower peak values. Landmarks are also more easily determined in able-bodied subjects than in some pathological gait patterns. For subjects with a total hip replacement prosthesis [16], peak muscle powers were not well defined which may impede the registration of the data. Nonetheless, clear peak power bursts were reported in below-knee amputees for the sagittal plane [17] as well as for other planes [18]. Also, Olney et al. [19] reported muscle power curves in hemiplegic patients which displayed a similar pattern allowing for identification of power bursts.

The major impact of applying curve registration to able-bodied gait data is to provide a representative mean curve where peak values have been aligned. These peak values taken from the registered curves correspond to the mean peaks calculated from the individual values taken on the unregistered curves. The registration technique is not intended to change the peak values of individual trials. Changes in the mean curve occur because the peak values do not occur at the same time. This technique would be appropriate when similar curves from different subjects are to be averaged.

Curve registration generally increased the peak muscle powers in all the joints of the lower limbs and each plane was about equally affected. It is interesting to note that the greatest increases occurred in the sagittal plane at push-off. The absolute increase in H3S, K3S and A2S totaled 1.00 W/kg by themselves, 0.80 W/kg of which were in power generation. Furthermore, S.D. of these three peak powers (Table 1) was reduced by curve registration.

Table 4
Mean and median frequency of the power spectrum (expressed in number of harmonics) and their respective standard deviations (S.D.) obtained for the unregistered and registered muscle power curves of the individual subjects[a]

Joint	Plane	Unregistered				Registered			
		Mean	S.D.	Median	S.D.	Mean	S.D.	Median	S.D.
Hip	Sagittal	1.63	0.16	3.44	0.27	1.59	0.14	3.37	0.24
	Frontal	2.31*	0.86	4.46	1.73	2.06*	0.54	3.89	1.02
	Transverse	2.04	0.78	3.98	1.42	2.14	0.71	4.03	1.18
Knee	Sagittal	1.64	0.36	3.18	0.71	1.65	0.35	3.37	0.56
	Frontal	2.11	0.48	4.06	1.19	2.17	0.54	4.22	1.30
	Transverse	2.46*	0.58	4.78	1.19	2.20*	0.38	4.45	0.68
Ankle	Sagittal	1.78	0.30	3.49	0.69	1.71	0.28	3.46	0.49
	Frontal	1.30*	0.43	2.34*	0.96	1.40*	0.43	2.45*	0.96
	Transverse	1.68	0.48	2.93	1.00	1.65	0.69	2.88	1.23

[a] Statistically significant differences are indicated by * for $P < 0.05$.

Following small shifts in the timing of the peaks, inflection points and zero-crossings by the registration process, the curve structural characteristics were not greatly modified. This is evidenced by the absence of any discontinuity in their first and second derivatives and by only small changes in the median or mean harmonics of their power spectra. However, the RMS values of the derivatives of the registered curves were larger than for the unregistered ones. This may be associated with the restoration of some of the characteristics of the original signals which were smoothed? out when the curves with unsynchronized peaks were grouped. While no suspect distortions in the registered curves were detected visually, no formal evaluation of this aspect was done. Consequently, part of the increase may also be associated with this factor. Like other types of normalization methods, the registration technique reduced inter-subject variability. It did not remove the inherent variability which exists between individual gait patterns. This is particularly important since push-off is closely associated with walking speed [2,20] and these values are often used to compare pathological performances with able-bodied gait.

Considering that curve registration has a tendency to slightly increase the peak powers and reduce the variability without substantially affecting the curve structural characteristics, it is recommended that curve registration be performed on able-bodied gait data prior to further statistical analyses. By reducing the S.D. in able-bodied gait parameters, the registration process would have a tendency to increase the probability of finding significant differences due to their smaller S.D.

5. Conclusion

Curve registration was responsible for a slight increase in peak muscle powers and a reduction in the variability of the muscle power curves. The mean increase was 0.10 W/kg in able-bodied subjects with the highest increases in peak muscle powers in the sagittal plane occurring at push-off. There was a 13.6% average decrease in the RMS value of the registered curves. Reductions were generally observed in the sagittal plane. No important discontinuities were reported in the first and second derivatives of the registered curves. Harmonic content of the power spectrum was not affected significantly. Curve registration is thus recommended prior to any gait pattern analysis of able-bodied subjects.

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