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Technical note

A measure of kinematic limb instability modulation by rhythmic auditory stimulation

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Abstract

A mathematical method based on computations of residual absolute value sums (RAVS) was developed for the quantitative analysis of tremor-like perturbations of knee angle during the gait cycle. The method was tested on simulation data created by adding sinusoidal tremor of varying frequency and amplitude to the knee-angle graph of a healthy test subject. The method was then applied to compare knee tremor reduction, with and without auditory rhythm, in a group of five traumatically brain-injured patients with gait hemiparesis. Deviations from normal gait performance due to tremor were assessed by using self-comparison to a 17th-degree regression polynomial of each subject's own motion-, time-, and point-normalized knee- angle curve. With rhythmic cueing, the five subjects had a statistically significant RAVS-measured mean tremor reduction of $39.5 \pm 22.6\%$ ($t = -3.91$; $p = 0.017$).

Keywords: Tremor; Gait; Rhythm; Knee position analysis; Polynomial regression

1. Introduction

This study proposes a method of analysis for lower extremity tremor based on the measure of residual absolute value sums (RAVS) of the knee angle versus time curve. The method was used to quantify tremor and the tremor reduction resulting from rhythmic cueing of walking, based on self-comparison to the 17th-degree regression polynomial of the knee joint angle versus time curve. Auditory rhythm has been previously shown to improve gait kinematic parameters in subjects with hemiparetic gait disorders (Prassas et al., 1997; Thaut et al., 1997, 1999).

Applications of traditional waveform analyses to tremor data, e.g. accelerometry with Fourier analysis (Frost, 1978), have shown limited usefulness for a variety of reasons, including the difficulty of standardizing movement-related tremor for quantification (Deuschl et al., 1991), misinterpretation of multiple irregular peaks in a tremor spectrum (Gresty and Buckwell, 1990), low correlation with other tremor measures, the tendency to overestimate tremor improvement (Bain et al., 1993), the necessity of matching the data-processing algorithm to the data characteristics (Timmer et al., 1996), and the absence of information relating to the amplitude and duration of tremor or the coincidence of tremor with well-defined stages of a movement sequence. Furthermore, while Fourier analysis can yield useful information about the frequency content of the tremor, in gait the positional perturbation of the knee angle has functional importance for the stability of the movement. Therefore, a waveform analysis showing tremor reduction based on positional data would yield information clinically and functionally more useful for gait assessment and therapy.

2. Methods

2.1. Normalization of data

Trial variations in walking speed were normalized by expressing time duration as a fraction of one complete gait cycle. To normalize for variations in the knee-angle range of motion of a given subject, the value of the minimum joint angle over the gait cycle was subtracted from each data angle,

$$\theta_{\min} = \min\{\theta_j\}_{j=1}^{N}, \quad (1)$$

$$\tilde{\theta}_i = \theta_i - \theta_{\min}, \quad i = 1, 2, \ldots, N \quad (2)$$

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and each shifted angle was then divided by the maximum shifted angle

$$\tilde{\theta}_{\max} = \max\{\tilde{\theta}_j\}_{j=1}^{N}, \tag{3}$$

$$\hat{\theta}_i = \frac{\tilde{\theta}_i}{\tilde{\theta}_{\max}}, \quad i = 1, 2, \ldots, N, \tag{4}$$

thus creating a dimensionless angle scale ranging in value from 0 to 1.

Each gait cycle was normalized to 1001 data points using an interpolation routine, since the RAVS procedure is dependent upon the number of points in the data sets. On the average, this increased the size of the data sets by a factor of four, and this level of interpolation has been shown to model human movement with finer resolution (Kenyon and Thaut, 2000).

2.2. Regression comparison reference

A tremor comparison reference was developed from a regressed version of each subject's personal knee-joint angle versus time curve, consistent with the assumption that tremor and other positional perturbations are superimposed on each subject's unique gait pattern. The choice of the degree for the regression polynomial was made on the basis of three procedures, each of which left the degree of the regression polynomial as a variable. First, the regression polynomial was tested for "goodness of fit" to a normal subject's knee-angle curve. The accuracy of the approximating regression polynomial was defined by Eq. (5), where ε represents the percent error measure, $\hat{\theta}$ is the normalized knee angle, and ρ is the corresponding knee angle after polynomial regression.

$$\varepsilon = \frac{\Sigma_{i=1}^{N} |\hat{\theta}_i - \rho_i|}{\Sigma_{i=1}^{N} |\hat{\theta}_i|} \times 100. \tag{5}$$

Second, it was required that the regression polynomial be a good approximation to the underlying non-perturbed gait patterns upon which tremor and other perturbations were superimposed. Third, the degree of the regression polynomial was "tuned" to give comparison results most consistent with the idealized changes within the tremor-simulated data sets, using tremor amplitude as the primary indicator of tremor severity.

2.3. Definition of the RAVS tremor measure M

The RAVS tremor measure M was defined as follows. First, a "residual" at a given data point was defined as the difference in value of the normalized knee angle and the regressed curve. The absolute values of the residuals were summed over the entire data set, then this sum was divided by the number of data points. In equation form

$$M = \frac{1}{N} \sum_{i=1}^{N} |\hat{\theta}_i - \rho_i| \tag{6}$$

where N = total number of normalized data points in the gait cycle. The resulting number may be thought of as the average tremor over the entire gait cycle. The percent change in tremor severity ΔT over two measurements (indicated by subscripts 1 and 2) was given by

$$\Delta T = \frac{M_2 - M_1}{M_1} \times 100. \tag{7}$$

2.4. Simulation

The validity of the RAVS measure M was tested by comparing changes in tremor severity, as calculated by Eq. (7), to tremor-amplitude changes in simulated data sets, which were constructed by mathematically adding sinusoidal tremor to the weight-bearing phase of one gait cycle of a 54-year-old male with normal gait. Ninety-seven simulations were created (including the zero tremor case) using tremor frequencies in the 4–15 Hz range (increments of 1 Hz), and tremor amplitudes in the 0.5–4.0° range (increments of 0.5°). The frequency range of 4–15 Hz includes the most widely documented results for tremor frequency, for a variety of tremor types, reported in the literature (Brooks et al., 1981; Deuschl et al., 1991; Elble et al., 1990; Frost, 1978; Ghika et al., 1993; Gresty and Buckwell, 1990; Jankovic and Frost, 1981; Marshall and Walsh, 1956; Timmer et al., 1996). The amplitude range, intended to represent barely noticeable to severe knee-joint tremors, was equal at its maximum to 6% of the normal subject's total knee-angle range, and included the tremor ranges of the five experimental subjects.

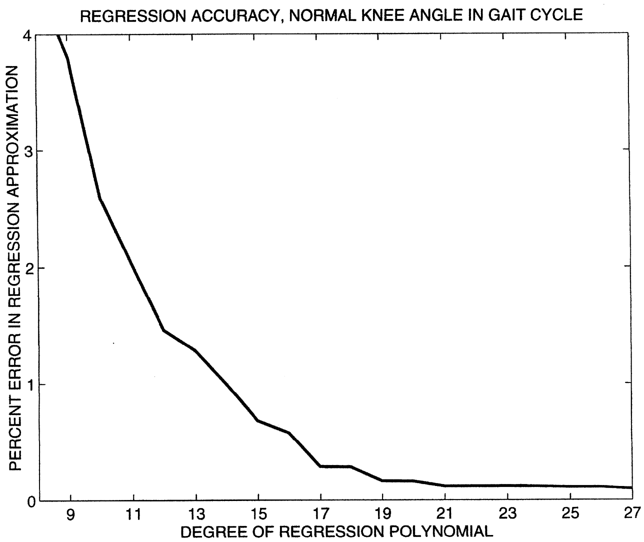

Fig. 1. This graph shows the error (as defined by Eq. (5)) in approximating the normal subject's knee angle over the gait cycle by a regression polynomial, with the degree of the regression polynomial as the independent variable. The approximation error is less than 1% when the degree of the regression polynomial is 15 or greater, and is monotonically decreasing.

Table 1
Summary of the results of applying the RAVS measure of tremor change to the 97 stimulated data sets (12 frequencies and 8 nonzero amplitudes, plus the zero amplitude case); (a) Shows tremor change calculated directly from amplitude change (Eq. (9)); (b) Shows mean tremor change over the entire frequency range as calculated by the RAVS method; (c) Shows the percent deviation of the Part (b) values from Part (a); (d) gives coefficients of variation for the percent deviations in Part (c).

A1	A2								
	0.0	0.5	1.0	1.5	2.0	2.5	3.0	3.5	4.0
(a) Ideals (percent amplitude change)									
0.5	− 100.0	0.0	100.0	200.0	300.0	400.0	500.0	600.0	700.0
1.0	− 100.0	− 50.0	0.0	50.0	100.0	150.0	200.0	250.0	300.0
1.5	− 100.0	− 66.7	− 33.3	0.0	33.3	66.0	100.0	133.3	166.7
2.0	− 100.0	− 75.0	− 50.0	− 25.0	0.0	25.0	50.0	75.0	100.0
2.5	− 100.0	− 80.0	− 60.0	− 40.0	− 20.0	0.0	20.0	40.0	60.0
3.0	− 100.0	− 83.3	− 66.7	− 50.0	− 33.3	− 17.0	0.0	16.7	33.3
3.5	− 100.0	− 85.7	− 71.4	− 57.1	− 42.9	− 28.6	− 14.3	0.0	14.3
4.0	− 100.0	− 87.5	− 75.0	− 62.5	− 50.0	− 37.5	− 25.0	− 12.5	0.0
(b) Mean tremor change by RAVS method									
0.5	− 67.6	0.0	95.7	192.3	288.8	385.0	480.9	576.3	671.3
1.0	− 83.4	− 48.9	0.0	49.3	98.6	147.7	196.6	245.4	293.9
1.5	− 88.9	− 65.8	− 33.0	0.0	33.0	65.9	98.7	131.3	163.8
2.0	− 91.6	− 74.2	− 49.6	− 24.8	0.0	24.7	49.4	73.9	98.3
2.5	− 93.3	− 79.4	− 59.6	− 39.7	− 19.8	0.0	19.8	39.4	59.0
3.0	− 94.4	− 82.8	− 66.3	− 49.7	− 33.1	− 16.5	0.0	16.4	32.8
3.5	− 95.2	− 85.2	− 71.0	− 56.8	− 42.5	− 28.3	− 14.1	0.0	14.0
4.0	− 95.8	− 87.0	− 74.6	− 62.1	− 49.6	− 37.1	− 24.7	− 12.3	0.0
(c) Percent deviation of means from ideals									
0.5	32.4	0.0	− 4.3	− 3.9	− 3.7	− 3.8	− 3.8	− 3.9	− 4.1
1.0	16.6	2.2	0.0	− 1.4	− 1.4	− 1.5	− 1.7	− 1.9	− 2.0
1.5	11.1	1.4	0.9	0.0	− 0.9	− 1.6	− 1.3	− 1.5	− 1.7
2.0	8.4	1.0	0.7	0.8	0.0	− 1.1	− 1.2	− 1.5	− 1.7
2.5	6.7	0.8	0.6	0.7	0.8	0.0	− 1.2	− 1.4	− 1.7
3.0	5.6	0.7	0.6	0.7	0.7	3.0	0.0	− 1.8	− 1.6
3.5	4.8	0.6	0.5	0.6	0.9	1.1	1.4	0.0	− 1.8
4.0	4.2	0.6	0.5	0.7	0.9	1.1	1.3	1.5	0.0
(d) Coefficients of variation									
0.5	2.9	0.0	4.9	4.8	4.8	4.9	4.9	5.1	5.2
1.0	1.4	2.5	0.0	2.4	2.4	2.5	2.6	2.8	2.9
1.5	0.9	1.7	1.6	0.0	1.7	1.8	1.9	2.1	2.3
2.0	0.7	1.3	1.2	1.3	0.0	1.5	1.7	1.9	2.0
2.5	0.6	1.0	1.0	1.1	1.2	0.0	1.6	1.7	1.9
3.0	0.5	0.9	0.9	1.0	1.1	1.3	0.0	1.7	1.9
3.5	0.4	0.7	0.8	0.9	1.0	1.3	1.4	0.0	1.8
4.0	0.4	0.7	0.8	0.9	1.0	1.2	1.4	1.6	0.0

It was assumed that, under conditions of constant tremor duration, tremor amplitude is a valid indicator of the functional severity of tremor, and that a change in tremor severity is best reflected by a change in tremor amplitude. Thus, for the purposes of the simulated data sets, percent change in tremor severity corresponded to the percent change in tremor amplitude, and was calculated as

$$\Delta A = \frac{A_2 - A_1}{A_1} \times 100. \tag{8}$$

The results for the change in tremor ΔT calculated by the RAVS method were then compared to the idealized amplitude changes ΔA.

2.5. *Pilot study with five subjects*

The RAVS tremor measure was applied to data collected from five subjects to assess tremor activity and changes in tremor activity under the cueing conditions of with and without RAS. Subjects were 3 right-hemiparetic and 2 left-hemiparetic patients with traumatic brain injury (TBI) (3 female, 2 male; mean age 32 ± 7 yr). Two subjects had mild, 1 had moderate, and 2 subjects had severe lower limb spasticity (Brunnstrom, 1970). All subjects had visually observable tremors only during the weight-bearing phase of the stride cycle. RAS was presented free-field as a metronome click (1000 Hz, 20 ms plateau), which was frequency-matched to the step frequency recorded and computed for the trial without RAS. One full gait cycle was recorded (using 60-Hz videocameras) for three-dimensional digitized video motion analysis.

3. Results

3.1. Regression polynomial

The degree of the regression polynomial was set at 17 because at this value, the regression polynomial converged to the normal subject's knee-angle curve with less than 1% error (Fig. 1), as defined by Eq. (5). It also converged to the normal knee angle upon which tremor had been superimposed in the 97 simulation data sets, with a mean error of $0.70 \pm 0.65\%$. Degree 17 also resulted in the overall lowest values for the percent deviations from the simulation results (Table 1).

3.2. Subject group

For the subject group, the RAVS-measured tremor was reduced by $39.5 \pm 22.6\%$ (Table 2) when RAS was applied. This change was statistically significant using dependent sample *t*-tests ($t = -3.914$; $p = 0.017$) as well as a non-parametric analysis using the Wilcoxon Signed Ranks Test ($Z = -2.023$; $p = 0.043$).

3.3. Illustrative example

Graphs for one subject (Fig. 2) show the regression polynomial and the actual knee–angle curve for the cases of with and without RAS. Without RAS, the RAVS tremor measure has a value of 5.99 (by Eq. (6)), and with RAS a value of 2.98. Thus the resulting change in tremor, as calculated by Eq. (7), is -50.3%, indicating a 50.3% decrease in tremor.

Table 2
Individual and group results for the RAVS tremor measure computed for gait data collected without (M_1) and with (M_2) a Rhythmic Auditory Stimulus (RAS) for subjects with traumatic brain injury (TBI). Tremors ranging from very moderate ($M_1 = 0.76$) to severe ($M_1 = 5.99$) showed significant improvement with RAS

Subject	M_1	M_2	Tremor change (%)
1	0.76	0.26	−66.1
2	5.12	4.80	−6.3
3	0.80	0.56	−30.1
4	2.55	1.41	−44.7
5	5.99	2.98	−50.3
Mean	3.04	2.00	−39.5
Standard deviation	2.42	1.88	22.6

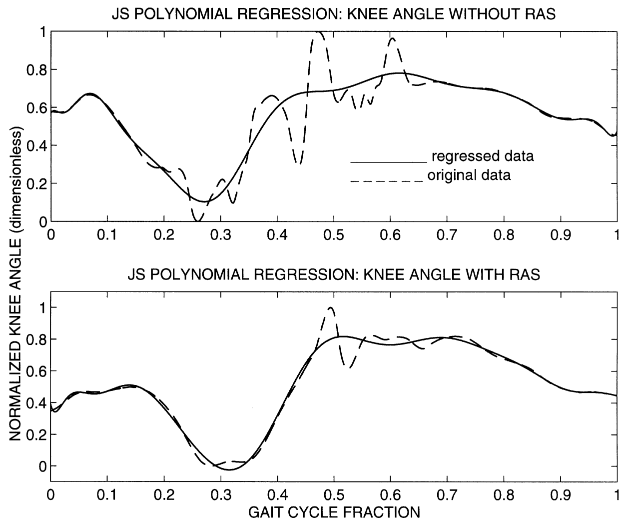

Fig. 2. For the two conditions of with and without rhythmic auditory stimulus (RAS), the dashed lines indicate the right knee angle over one complete gait cycle and the solid lines are the corresponding degree 17 regression polynomials. The RAVS method in essence measures the change, from one case to another, of the magnitude of the areas between the solid and dashed lines.

4. Discussion

The RAVS method applied to our study sample quantified and confirmed that tremor was reduced during rhythmic cueing. Considering motor control as an optimization problem of spatiotemporal precision (Harris and Wolpert, 1999), trajectory smoothing during RAS may be explained as a consequence of adding temporal stability and precision to the movement through the entrainment of the motor response to the rhythmic timekeeper.

The RAVS method, derived as a mathematical tool to quantify tremor severity, showed good results in computing tremor strength and changes in tremor as well as being sensitive enough to analyze degrees of tremor severity from very mild to very pronounced. Quantification of tremor change is an essential and new feature of this method, since no measure of tremor reduction has been found in the literature. Furthermore, unlike acceleration-based analyses, RAVS analysis uses position data itself rather than data derived from position, which allows analysis of duration and amplitude of tremor, and of correlations of tremor activity with movement landmarks of functional significance.

From a purely theoretical point of view, it is noteworthy that a direct mathematical correspondence exists between the amplitude of pure sinusoidal tremor and its RAVS-measure. For a tremor of frequency f Hz and amplitude A, with Δt as the time interval between RAVS data points, it can be shown, by relating both tremor amplitude and the RAVS measure of tremor to the area contained by the tremor curve, that the RAVS measure is proportional to the tremor amplitude:

$$M = \frac{2A}{N\pi f \Delta t}. \quad (9)$$

This result suggests that changes in tremor ΔT computed by the RAVS method should correspond closely with changes in tremor amplitude, assuming constant duration and location of tremor (i.e., weight-bearing phase) over trials.

The RAVS method may be particularly useful when analyzing pathological gait, because comparisons are made to each subject as their own "normalized" reference instead of comparing pathological gait patterns to averaged gait norms established with other study groups. Although the patient group of this study had tremor confined to the stance phase of the gait cycle, the RAVS method was also tested against simulation data sets with variable tremor duration over the entire gait cycle, with similar results as for the variable-amplitude simulations.

References

Bain, P.G., Mally, J., Gresty, M., Findley, L.J., 1993. Assessing the impact of essential tremor on upper limb function. Journal of Neurology 241, 54–61.

Brooks, V.B., Brookhart, J.M., Mountcastle, V.B., 1981. Motor control. In: Geiger, S.R. (Ed.), Handbook of Physiology, Vol. II. American Physiology Society, Bethesda, MD, pp. 325–343.

Brunnstrom, S., 1970. Movement Therapy in Hemiplegia. Harper & Row, New York, pp. 343–344.

Deuschl, G., Blumberg, H., Lucking, C.H., 1991. Tremor in reflex sympathetic dystrophy. Archives Neurology 48, 1247–1252.

Elble, R.J., Sinha, R., Higgins, C., 1990. Quantification of tremor with a digitizing tablet. Journal of Neuroscience Methods 32 (3), 193–198.

Frost, J.D., 1978. Triaxial vector accelerometry: a method for quantifying tremor and ataxia. IEEE Transactions on Biomedical Engineering BME-25, 17–27.

Ghika, J., Wiegner, A.W., Fang, J.J., Davies, L., Young, R.R., Growdon, J.H., 1993. Portable system for quantifying motor abnormalities in Parkinson's Disease. IEEE Transactions on Biomedical Engineering 40, 276–283.

Gresty, M., Buckwell, D., 1990. Spectral analysis of tremor: understanding the results. Journal of Neurology, Neurosurgery and Psychiatry 53, 976–981.

Harris, C.M., Wolpert, D.M., 1999. Signal-dependent noise determines motor planning. Nature 394, 780–784.

Jankovic, J., Frost, J.D., 1981. Quantitative assessment of parkinsonian and essential tremor: Clinical application of triaxial accelerometry. Neurology 31, 1235–1240.

Kenyon, G.P., Thaut, M.H., 2000. The discretization and interpolation of human movement and their effects on reported acceleration. Submitted for publication.

Marshall, J., Walsh, E.G., 1956. Physiological temor. Journal of Neurology, Neurosurgery and Psychiatry 19, 260–267.

Prassas, S.G., Thaut, M.H., McIntosh, G.C., Rice, R.R., 1997. Effect of auditory rhythmic cuing on gait kinematic parameters of stroke patients. Gait and Posture 6, 218–223.

Thaut, M.H., Kenyon, G.P., Schauer, M.L., McIntosh, G.C., 1999. The connection between rhythmicity in brain function: implications for therapy of movement disorders. IEEE Engineering, Medicine and Biology 18, 101–108.

Thaut, M.H., Rice, R.R., McIntosh, G.C., 1997. Rhythmic facilitation of gait training in hemiparetic stroke rehabilitation. Journal of Neurological Sciences 151, 207–212.

Timmer, J., Lauk, M., Deuschl, G., 1996. Quantitative analysis of tremor time series. Electroencephalography Clinical Neurophysiology 101, 461–468.