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# Some Methods and Parameters of Body Sway Quantification and Their Neurological Applications*

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**Summary.** Methods and parameters are described to quantify body sway as measured by a force-transducing platform. Analogue data representing the coordinates of the body's center of force (COF) are fed into a digital computer. The following parameters are then calculated and tested for their diagnostic significance: sway path (SP), mean amplitude of sway (MA), mean sway frequency (MF), their lateral and sagittal components, and the quotients sagittal/lateral of these as well as the sway area (SA) circumscribed by the COF. Quotients of eyes open/eyes closed for all these parameters determine the visual stabilization of posture. Sway position and sway direction histograms allow for a more detailed analysis of MA and SP. Despite considerable inter- and intraindividual variance of these parameters (in 28 normals), some of them seem of clinical significance not only for documentation and follow-up studies but also for differential diagnosis.

In patients with cerebellar lesions ($n = 12$), SP and MA were up to 10 times larger with a marked antero-posterior instability, MF being above normal. Patients with labyrinthine lesions ($n = 10$) showed significant instability only with eyes closed, MF being slightly below normal.

**Key words:** Posture – Quantitative evaluation – Cerebellum.

**Zusammenfassung.** Es werden Methodik und Parameter für eine quantitative Auswertung der Standunruhe bei Kranken und Gesunden beschrieben. Aus den Analogsignalen einer Kraftmeßplattform werden die folgenden Daten über die Bewegungen des Körperkraftschwerpunktes digital errechnet: Schwingungsweg (SP), dessen mittlere Amplitude (MA), mittlere Frequenz (MF), die lateralen und sagittalen Komponenten, deren Quotienten und die von den Körperschwankungen umschlossene Fläche (SA). Quotienten der

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Werte bei offenen und geschlossenen Augen beschreiben die visuelle Standstabilisation. Positions- und Richtungshistogramme dokumentieren zusätzlich die Vorzugsrichtungen der Körperunruhe.

Eine Analyse der Signifikanz dieser Parameter zeigte trotz großer Varianz bei Normalen deren klinische Brauchbarkeit. Patienten mit *Kleinhirnerkrankungen* (vorwiegend des Vorderlappens) zeigen bis zu 10mal höhere Werte, z. B. für MA und SP vorwiegend in sagittaler Richtung und eine erhöhte mittlere Frequenz (MF), während Patienten mit weitgehend kompensierten *Vestibularisläsionen* nur bei geschlossenen Augen vermehrt schwanken, wobei MF erniedrigt ist.

**Schlüsselwörter:** Standstabilität – Quantitative Auswertung – Kleinhirn.

## Introduction

The upright stance of the human is an unstable position. Even normals exhibit a rather small postural sway which reflects noise and regulatory activity of the several control loops involved in maintenance of balance. Balance requires that the body's center of gravity never deviates beyond the support area. The main sensory systems involved are proprioception, the vestibular system and vision, and their afferent pathways within the central nervous system. Afferent and efferent pathways involve the spinal cord, the brain stem, the cerebellum, the midbrain, and the sensorimotor cortex. Postural instability consequently may result from lesions within any part of the stabilizing systems. By quantitative analysis it may also be investigated whether characteristically altered postural sway parameters can be correlated to the location of a lesion within the complicated multiloop system. Furthermore, clinical analysis of ataxia aims at quantitative documentation of postural sway for follow-up studies.

Several recording techniques have been reported since the mechanical devices of Miles (1922) and Edwards (1941): measurements of the distance-dependent capacity between the subject and an external metallic reference (Uchytil, 1962; Lee and Lishman, 1975), photographic registration of the sway path of a light fixed to the subject's head (Monserrat, 1969; Claussen, 1970), and the most modern techniques of recording the displacement of the center of force on a force-measuring platform (Hellebrandt, 1938; Bensel and Dzendolet, 1968; Kapteyn, 1972; Dichgans et al., 1976). Simple sway parameters were defined and tested by Kapteyn (1972), Hirashawa (1973), Njiokiktjien and van Parys (1976), and most recently by Taguchi et al. (1978). Among the parametric methods of analysis, the analysis of the Fourier power spectrum has been shown to be most effective (Dichgans et al., 1976; Mauritz et al., 1979).

This paper reports a number of new parameters quantifying postural sway in normals and patients with cerebellar and labyrinthine lesions as well as their variance, reliability, and clinical significance.

## Methods

### *Measurement of Body Sway*

Body sway may be recorded in terms of the displacement of the center of force (COF) generated by the inherent instability of a subject standing on a recording platform. The center of force is not identical to the center of gravity since it entails dynamic forces (see below).

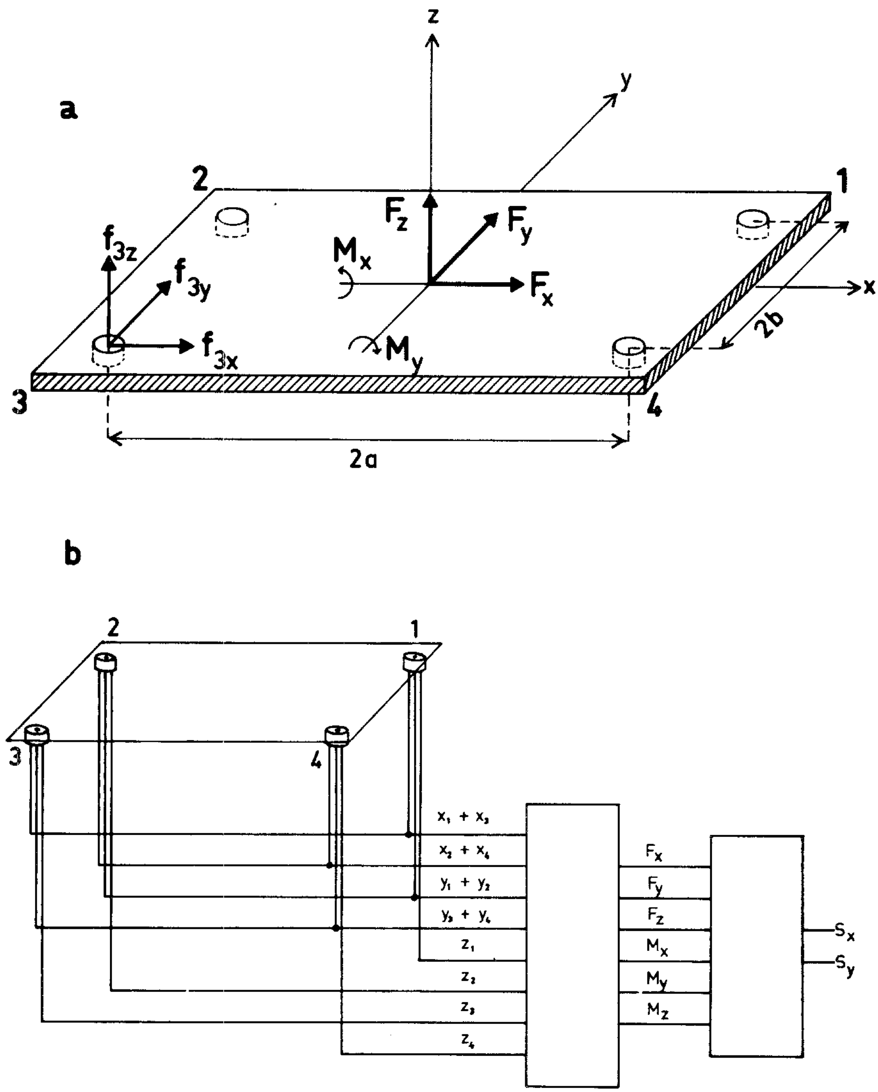


**Fig. 1. a** Component forces and force moments acting on the platform; **b** schematic presentation of the combination of component forces and further analogue computation in two steps

The platform (Kistler) is mounted on four piezoelectric force transducers ($i = 1 \ldots 4$), each of which measures forces in the three main directions: $f_{ix}, f_{iy}, f_{iz}$ (Fig. 1a). Signals are combined as indicated in Figure 1b. To obtain the resultant forces and force movements acting on the platform along the axes $x$, $y$, and $z$, the components are added as follows:

$$\vec{F} = \sum_{i=1}^{4} f_i = \begin{pmatrix} f_{1x} + f_{2x} + f_{3x} + f_{4x} \\ f_{1y} + f_{2y} + f_{3y} + f_{4y} \\ f_{1z} + f_{2z} + f_{3z} + f_{4z} \end{pmatrix} \qquad (1)$$

$$\vec{M} = \sum_{i=1}^{4} \vec{m}_i = \sum_{i=1}^{4} \vec{r}_i \times f_i$$

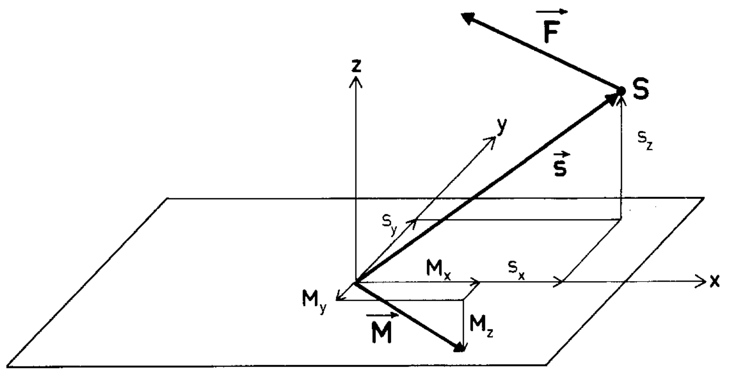


**Fig. 2.** Forces and force moments acting on the platform with a subject standing on it. The subject's center of gravity in *S*. *M* is a force moment applied to the platform by an eccentric force *F* acting on *S*. Horizontal coordinates $s_x$ and $s_y$ of *S* are calculated from the component forces and force moments recorded by the platform

$r_i$ are the position vectors of the force transducers. In a second analogue-computing unit, the final output signals $s_x$ and $s_y$ are calculated. They define the momentary position of the center of force (COF) on the platform, the motion of which describes the sway path. The physical definition of this point requires the following consideration. The subject stands on the platform with his center of gravity (COG) in *S* (Fig. 2). $\vec{s}$ in Figure 2 is the position vector of *S*:

$$\vec{s} = \begin{pmatrix} s_x \\ s_y \\ s_z \end{pmatrix} \tag{2}$$

Under certain admittedly simplified assumptions, the total mass of the body can be treated as if it were concentrated in *S* and attached to the platform at the end of a weightless lever like an inverted pendulum. These assumptions are: (a) that there are no segmental movements between parts of the body and (b) that all the forces acting on the body act on *S*. A force $\vec{F}(f_x, f_y, f_z)$ acting on *S* creates a force moment $\vec{M}$ in the middle on the platform which is equal to the vector product of *s* and *F*:

$$\vec{M} = \vec{s} \times \vec{F} = \begin{pmatrix} s_yF_z - s_zF_y \\ s_zF_x - s_xF_z \\ s_xF_y - s_yF_x \end{pmatrix} \tag{3}$$

The component forces of $\vec{F}$ and *M*, as calculated by the first analogue-computing unit, are combined according to equation 4 which is derived from equation 3 to obtain $s_x$ and $s_y$:

$$s_x = \frac{s_zF_x - M_y}{F_z} \qquad s_y = \frac{s_zF_y + M_x}{F_z} \tag{4}$$

*s* is set to a constant height of 1 m.

In reality, the body is not rigid and active muscle forces are at work in it. Therefore, the definition given above is not entirely correct. The coordinates of *S* closely match the coordinates of the actual center of gravity only in slow motions (low frequencies). At increasing frequencies, dynamic components due to body inertia progressively add to the static forces and account for 50% at 0.5 Hz, whereas they measure less than 10% at 0.2 Hz (Gurfinkel, 1973). Our recording technique, therefore, measures not only the net displacement of the COG but also inertial forces and force movements created by the torque about the ankles. Consequently the vertical

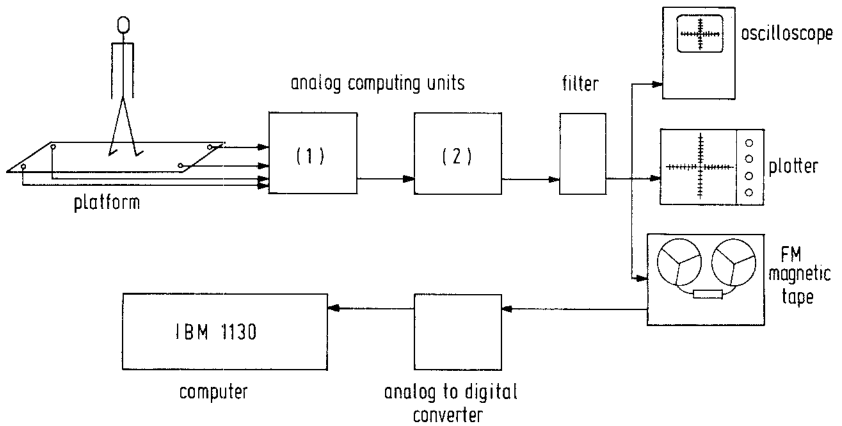


**Fig. 3.** Principal parts of the measuring device

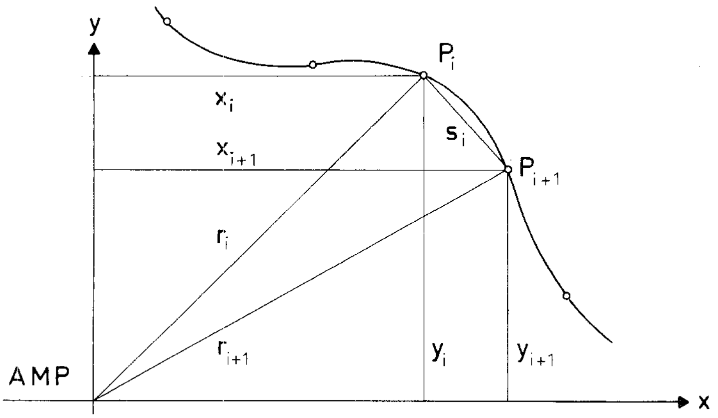


**Fig. 4.** Diagrammatic presentation of the sway parameters: mean amplitude *(MA)* = mean of distances $r_i$; sway path *(SP)* = sum of $s_i$; sway area *(SA)* = sum of the area of triangles AMP $P_i$ $P_{i+1}$

projection of $S$ with coordinates $s_x$ and $s_y$ is called the center of force (COF) and not the center of gravity (COG).

The output signals $s_x$ and $s_y$ of the second analogue-computing unit are then passed through a low-pass filter damping frequencies above 40 Hz (damping ratio +20 db/decade). Signals are displayed on an oscilloscope or plotter and stored on FM magnetic tape (Fig. 3).

Output signals $s_x$ and $s_y$ were calibrated under static conditions using weights placed at different distances and in different directions on the platform.

### *Quantification of Visual Influence Upon Posture*

The stabilizing effect of vision was assessed by comparing the measurements of sway while standing with eyes open and eyes closed. The destabilizing effect of large moving visual scenes upon posture (Dichgans et al., 1972, 1976) was studied using a dome rotating about the subject's line of sight at an angular velocity of 50°/s. The dome covered the entire visual field. The inner surface of the dome was covered with a random dot pattern.

### *Quantitative Sway Parameters*

$s_x$ and $s_y$ signals were converted from analogue to digital with a sampling interval of 30 ms and a total sampling duration of 37 s. The third channel of the analogue-to-digital converter was used to sample the noise of an empty tape track in order to later correct for its contribution to our measurements. Thus, a total number of 1299 single values was obtained per channel. High frequency tape noise was eliminated by passing the $x$ and $y$ values through a digital Butterworth low-pass filter with an edge frequency of 4 Hz (subprogram by J. H. J. Allum).

It was not possible to place the different subjects so that the average vertical projection of their center of gravity on the platform was identical even if they had the same preset position of the feet. Therefore, to normalize the data, the *arithmetic mean point (AMP)* with the means of all the $s_x$ and $s_y$ values within one 37-s measuring period as coordinates was calculated for each experiment. All the other parameters were related to the AMP.

The following parameters were computed (Fig. 4):

*mean amplitude (MA),* i.e., the mean distance between the sampling points and the AMP:

$$\mathrm{MA} = \sum_{i=1}^{n} r_i = \sum_{i=1}^{n} \sqrt{x_i^2 + y_i^2} \tag{5}$$

*sway path (SP),* i.e., the length of the path described by the COF in one minute, which is approximated by the sum of the distances between two consecutive sampling points:

$$\mathrm{SP} = \sum_{i=1}^{n-1} s_i = \sum_{i=1}^{n-1} \sqrt{(x_{i+1} - x_i)^2 + (y_{i+1} - y_i)^2} \tag{6}$$

*mean frequency (MF),* i.e., the frequency of a circular motion with a radius equal to MA and a total length identical to the sway path (Fig. 5a):

$$\mathrm{MF} = \frac{\mathrm{SP}}{2\pi\,\mathrm{MA}} \tag{7}$$

MF in fact gives a rough measure of the mean sway frequency favoring high-frequency components due to the over-proportional contribution of fast motions to COF sway which are not accompanied by adequate shifts of the COG.

*lateral and sagittal components* of the three afore-mentioned parameters ($\mathrm{MA}_{lat}$, $\mathrm{MA}_{a\text{-}p}$, $\mathrm{SP}_{lat}$, $\mathrm{SP}_{a\text{-}p}$, $\mathrm{MF}_{lat}$, $\mathrm{MF}_{a\text{-}p}$). The mean frequency of a direction component $\mathrm{MF}_{comp}$ is defined as the frequency $f$ of a sinusoidal oscillation, the sway path and amplitude of which are identical to the corresponding parameters of body sway in this direction (Fig. 5b):

$$f = \frac{\bar{v}}{4 \cdot s_0} \tag{8}$$

$$s_0 = \bar{s} \cdot \sqrt{2} \tag{9}$$

$\bar{v}$ = mean velocity of the sinusoidal oscillation
$s_0$ = amplitude.
Consequently, the mean frequency of one sway component is defined by

$$\mathrm{MF}_{comp} = \frac{\mathrm{SP}_{comp}}{4 \cdot \sqrt{2} \cdot \mathrm{MA}_{comp}} \tag{10}$$

*sway area (SA),* i.e., the total area circumscribed by the COF within one minute. This was approximated by summing up the area of the triangles defined by AMP, $P_i$ and $P_{i+1}$ in Figure 4.

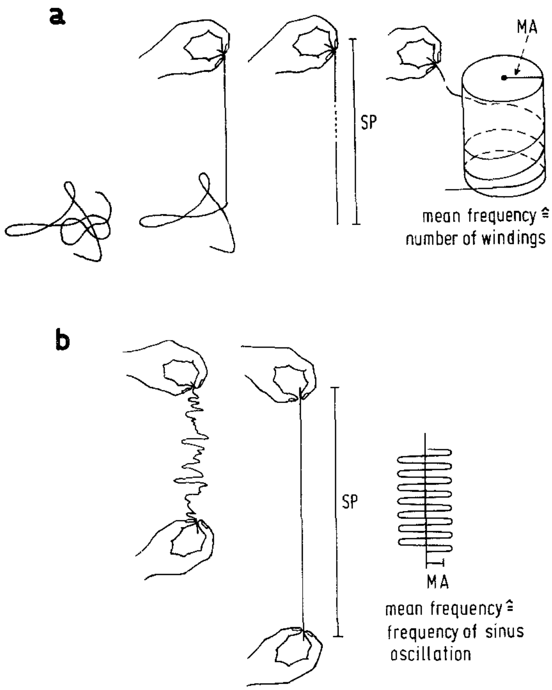


**Fig. 5 a and b.** Mean frequency *(MF)* illustrated by a simple model. The mean amplitude *(MA)* is symbolized by a cylinder with a radius MA; the sway path *(SP)* of 1 s by a rope of identical length. When SP is wound around the cylinder, the mean frequency corresponds to the number of windings (**a**). A similar procedure is used to visualize the mean frequency of a sway component (**b**): the rope is formed into a sinusoid with a mean amplitude MA. MF corresponds to the number of cycles

The sway area is a mixed parameter combining sway amplitude and sway path, the latter contributing a greater amount.

*quotients of sagittal/lateral component* of mean amplitude (QMA), sway path (QSP), and mean frequency (QMF). Differences between the two components were also tested but turned out to be less reproducible than the quotients.

*quotients of eyes closed/eyes open* ($Q_{vis}$, 'Romberg quotient') of all the afore-mentioned parameters showing the amount of visual stabilization of posture.

*quotients of sway with roll vection stimulus/eyes open without stimulus* ($Q_{stim}$) of all parameters assessing the amount of visual destabilization by the stimulus.

To compensate for tape noise which invariably causes overestimations, especially of the sway path, the bias due to tape noise and flutter is computed from the third tape track for all principal parameters and is then subtracted. Calculation of the mean frequency and the quotients is based on these corrected values.

The following considerations regarding the physiologic significance of the computed parameters should be taken into account. In a subject standing quietly, the AMP can be regarded as a set value and sway, the control system's continuous attempt to match the COF with the AMP. The mean amplitude gives the mean distance between the COF (the actual locus) and the AMP (the set value) and represents a reciprocal parameter of the effectiveness of balance or, in other words, a parameter of instability. Among the parameters defined, the sway path best describes the regulatory activity of the balancing systems, while it is a poor measure of the passive shifts of

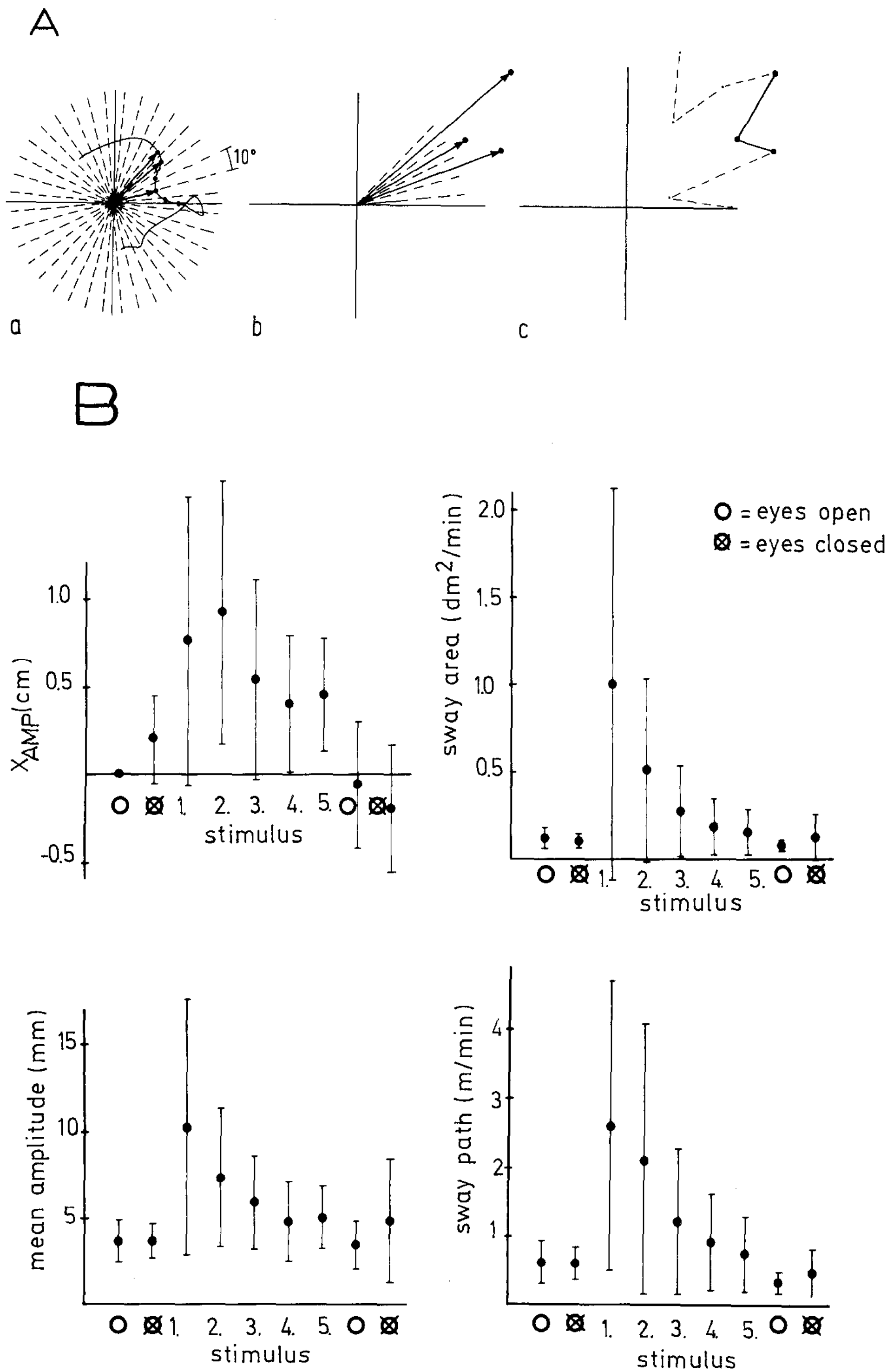


**Fig. 6. A)** Schematic presentation of angle histogram computation. The amounts of the position vectors $r_i$ are summed up within angle intervals of 10° *(a)*, the sum is depicted as a new vector in the middle of the interval *(b)*, each indicating one point of the position histogram *(c)*. **B)** Habituation effects under repeated visual stimulation (means of 5 normal subjects)

the COG. This is because the sway path entails all the dynamic forces which become exceedingly preponderant in the high frequency range of the power spectrum. The contribution of passive motions (falling) is rather small.

*Sway Angle Histograms*

Simple quantitative parameters indicating the predominance of principal sway directions have already been given by quotients QMA and QSP. To obtain a better idea of the positional and directional distribution of postural sway, the following two kinds of graphical presentations were tested. To this effect, the full circle of possible directions is divided into 36 intervals of 10° width each. A *position histogram* is constructed from the position vectors of all sampling points (Fig. 6Aa). According to its polar angle, each of the position vectors falls into one of the 36 intervals. The lengths of all the position vectors within one interval are added. Thus a new vector is constructed, with its polar angle in the middle of the interval and its length equal to the sum (Fig. 6Ab). Linking the points given by these 'sum' vectors, we obtain the position histogram (Fig. 6Ac). Summing up tha path vectors $(\overrightarrow{P_i P_{i+1}})$ instead of the position vectors, we get the *sway direction histogram,* i.e., the angular distribution of sway path.

## Results

### *Data from Normals*

Table 1 gives the average data of the most valuable parameters from normals. Parameters which later on turned out to be less useful for clinical purposes $(Q_{vis}, Q_{stim})$ were excluded.

Eye closure produces a significant increase of sway amplitude, path, and area, especially in the *a-p* direction. The relative increase of the sway path (balancing activity) is larger than that of the mean amplitude (postural instability). This can, in comparison to some of the patient data quoted below, be interpreted as the consequence of a fairly effective postural regulation. Visual destabilization of posture by a rotating visual field, despite great interindividual differences, invariably shifts the COG (AMP) toward the direction of rotation and mainly induces sway in the lateral direction. This may be seen from quotients QMA and QSP. When exposed to repeated stimulation, subjects showed habituation (Fig. 6B).

The decrease of AMP displacement with an increasing number of stimulations was accompanied by a decrease in all of the sway parameters. It might be concluded that the 'sensory weight' of visual cues in postural regulation is adaptable and decreases with long-lasting conflicting stimuli. The quotients eyes closed/eyes open $(Q_{vis})$ are, however, not significantly lower after repeated stimulation (Table 2), which indicates a lack of transfer of this habituation to normal visual stabilization. The reason for this may be that the rotation velocity of the dome, 50°/s, is much higher than the peak angular velocity of retinal image motion reached under physiologic conditions. A peak velocity of 50°/s at a maximum sway amplitude of 5° would correspond to a sway frequency of 1.6 Hz. Normal sway, however, has its greatest power in the Fourier spectrum at frequencies well below 1 Hz, and higher frequency components are mainly produced by dynamic forces and not by shifts of the COG. Higher frequency components, therefore, do not cause relevant shifts in relation to the visual

**Table 1.** Means and standard deviations of the most valuable parameters ($n$ = 28)

| | Eyes open | Eyes closed | Visual stimulus |
|---|---|---|---|
| MA (mm) | 3.44 (1.40) | 4.75 (1.65) | 9.22 (4.74) |
| $MA_{lat}$ (mm) | 1.74 (0.74) | 2.51 (1.12) | 6.59 (4.10) |
| $MA_{a\text{-}p}$ (mm) | 2.65 (1.17) | 3.52 (1.25) | 5.14 (2.21) |
| QMA | 1.61 (0.60) | 1.63 (0.88) | 0.95 (0.34) |
| SP (m/min) | 0.64 (0.28) | 1.01 (0.45) | 2.38 (1.65) |
| $SP_{lat}$ (m/min) | 0.29 (0.16) | 0.50 (0.29) | 1.65 (1.39) |
| $SP_{a\text{-}p}$ (m/min) | 0.50 (0.23) | 0.76 (0.34) | 1.36 (0.76) |
| QSP | 1.94 (0.85) | 2.07 (1.56) | 1.06 (0.37) |
| MF (Hz) | 0.51 (0.21) | 0.56 (0.18) | 0.65 (0.22) |
| $MF_{lat}$ (Hz) | 0.50 (0.24) | 0.57 (0.24) | 0.69 (0.28) |
| $MF_{a\text{-}p}$ (Hz) | 0.62 (0.30) | 0.63 (0.22) | 0.76 (0.25) |
| QMF | 1.39 (0.87) | 1.28 (0.61) | 1.20 (0.42) |
| SA ($dm^2$/min) | 0.09 (0.06) | 0.16 (0.11) | 0.82 (0.85) |

**Table 2.** Quotients eyes closed/eyes open before and after repeated visual stimulation (means of 8 normal subjects)

| | Before stimulation | After stimulation |
|---|---|---|
| MA | 1.04 | 1.14 |
| $MA_{lat}$ | 1.16 | 1.24 |
| $MA_{a\text{-}p}$ | 0.98 | 1.09 |
| SP | 1.24 | 1.19 |
| $SP_{lat}$ | 1.38 | 1.17 |
| $SP_{a\text{-}p}$ | 1.17 | 1.21 |
| SA | 1.09 | 1.09 |

**Table 3.** Methodological variance of the parameters. Standard deviation *(left)* and standard deviation in per cent of the normal value *(right)*

| | $s$ | $\frac{s}{\text{normal value}} \cdot 100$ |
|---|---|---|
| MA (mm) | 0.114 | 3.3 |
| SP (m/min) | 0.064 | 10.0 |
| MF (Hz) | 0.018 | 3.5 |
| SA ($dm^2$/min) | 0.008 | 8.9 |

| | Eyes open $v$ (%) | Eyes closed $v$ (%) | Stimulus $v$ (%) |
|---|---|---|---|
| MA | 33.5 | 23.7 | 33.9 |
| SP | 24.6 | 32.9 | 40.6 |
| MF | 24.6 | 30.1 | 21.9 |
| SA | 58.9 | 47.5 | 60.4 |

**Table 4.** Variation coefficients ($v = \frac{s}{x} \cdot 100$) from several (up to 11) tests with the same subject. Mean values from 8 normal subjects

| | Eyes open $v$ (%) | Eyes closed $v$ (%) | Stimulus $v$ (%) |
|---|---|---|---|
| MA | 40.5 | 34.7 | 51.3 |
| SP | 43.8 | 44.4 | 69.1 |
| MF | 39.6 | 32.5 | 34.2 |
| SA | 57.8 | 64.4 | 102.3 |

**Table 5.** Variation coefficients ($v = \frac{s}{x} \cdot 100$) of interindividual variance between 28 normal subjects

surroundings. Moreover, visual stabilization of normal stance, according to Nashner (1970) and Dichgans et al. (1976), is most effective below 0.2 Hz, so the decrease of sensory weight may be velocity-specific.

*Methodological, Intra- and Interindividual Variance of the Parameters*

The methodological variance was investigated by repeatedly evaluating the same test interval of 37 s from the tape. This was done ten times each after new calibration.

The standard deviation of the principal parameters thus obtained is shown in Table 3. The variance is due to calibration errors and additive errors in the estimate of noise. The latter component is responsible for the fact that the relative methodological dispersion decreases in high sway values, making pathologic results more precise than normal ones. Intraindividual variance was studied by making several tests (up to 11) with the same subjects. Table 4 shows the surprisingly high average variation coefficients of eight normal subjects. Variance was greater in subjects with rather small sway amplitudes. Table 5 shows variation coefficients of interindividual variance. These, quite expectedly, even exceed those of intraindividual variance. The total number of subjects tested was 28.

*Patient Data*

*Cerebellar Ataxias.* 12 patients with cerebellar ataxias of various origins, on average showed a significantly larger sway amplitude (MA) and sway path (SP) than normals and, consequently, a high mean frequency (MF; Fig. 7A). In cases

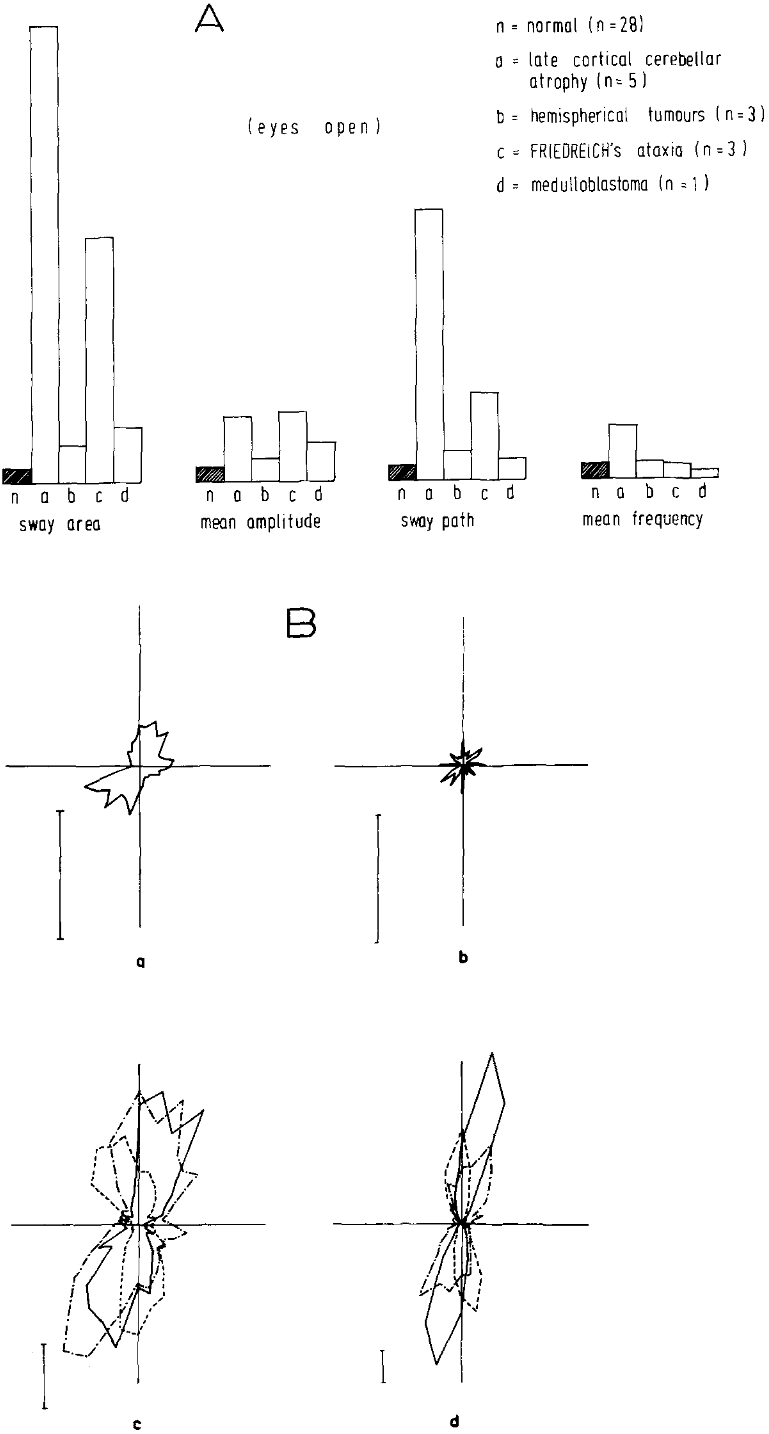


**Fig. 7. A)** Quantitative sway parameters of 12 patients with cerebellar ataxia. **B)** Position *(a)* and direction histogram *(b)* of one normal person *(a, b)* compared to histograms of three typical cerebellar atrophy patients *(c, d)*. Large predominance of *a-p* displacement *(c)* and motion direction *(d)*. The notch at the *x*-axis in *(d)* is due to superposition of sagittal motions on all lateral motions and, therefore, lack of purely lateral sway

**Table 6.** Number of pathologic results (parameters above the 2s range) in patients with various cerebellar diseases

| | Late cortical cerebellar atrophies ($n$ = 5) | | Hemispherical tumors ($n$ = 3) | | Friedreich's ataxia ($n$ = 3) | | Medullo-blastoma ($n$ = 1) | |
|---|---|---|---|---|---|---|---|---|
| | e.o. | e.c. | e.o. | e.c. | e.o. | e.c. | e.o. | e.c. |
| MA | 4 | 5 | 1 | 2 | 3 | 3 | 1 | 0 |
| $MA_{lat}$ | 4 | 5 | 1 | 2 | 3 | 3 | 1 | 0 |
| $MA_{a\text{-}p}$ | 4 | 5 | 0 | 2 | 3 | 3 | 1 | 0 |
| QMA | 1 | 1 | 0 | 0 | 0 | 0 | 0 | 0 |
| SP | 5 | 5 | 1 | 1 | 3 | 3 | 0 | 0 |
| $SP_{lat}$ | 4 | 5 | 1 | 2 | 3 | 3 | 0 | 1 |
| $SP_{a\text{-}p}$ | 5 | 5 | 2 | 2 | 3 | 3 | 0 | 0 |
| QSP | 1 | 2 | 0 | 0 | 0 | 0 | 0 | 0 |
| SA | 4 | 5 | 1 | 2 | 3 | 3 | 1 | 0 |

e.o. = eyes open
e.c. = eyes closed

with late cortical cerebellar atrophy of the anterior lobe and in patients with Friedreich's ataxia, MA, Sp, their *a-p* and lateral components, and SA were considerably larger and nearly without exception fell outside the 2s range of normality (Table 6). Neocerebellar tumors, however, although on the average above the mean of normals, were more often within the 2s range.

The predominant increase of sway path (balancing activity) and less increase of mean amplitude (instability) as well as a 3 Hz peak of *a-p* sway, characteristic for patients with anterior lobe atrophy (Dichgans et al., 1976; Mauritz et al., 1979) reflect an underdamped action of the balancing systems, in contrast to the predominant increase of instability found in the labyrinthine type of postural ataxia. A further characteristic of cerebellar ataxias, again mainly those due to anterior lobe atrophy, is the predominance of sway in the *a-p* direction (Fig. 7B). This may well be demonstrated and documented by the sway position and sway direction histograms and is not seen with any other type of postural ataxia so far investigated.

According to the results presented in Table 6, the principal parameters are more useful for diagnosis than the quotients, although the total sway pattern can only be characterized by a combination of several parameters. The most sensitive parameter seems to be the sway area (SA).

*Labyrinthine Lesions.* Ten patients were tested. The age of the lesion varied between one day and two years. All patients had no or a decreased caloric response of one labyrinth. Some patients showed a visible inclination towards the side of the lesion. This observation was quantified by comparing the average

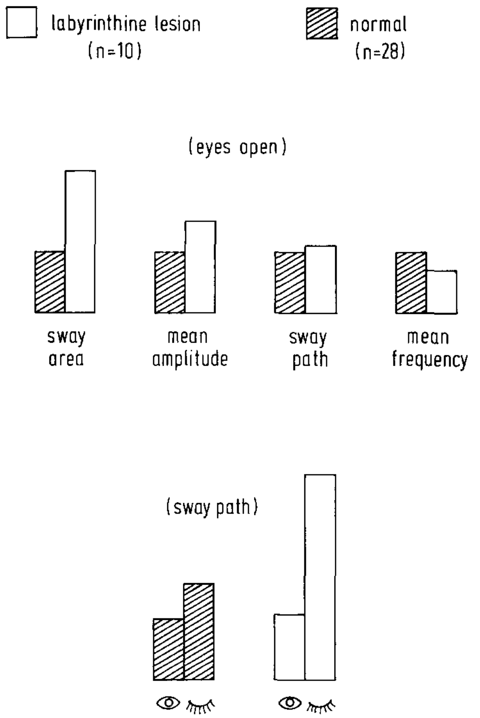


**Fig. 8.** Quantitative sway parameters of 10 patients with labyrinthine lesions. Parameters are not significantly higher with eyes open, but definitely increased with eyes closed

COF position (AMP) with eyes open and eyes closed. The average deviation was 8 mm ($P < 0.02$). Sway parameters on the average were above normal with eyes open but were significantly larger only in little more than half of the cases (MA: 60%, SP: 50%, SA: 60%). The other parameters were even less discriminative. The pattern changed toward a significant increase of instability when the eyes were closed (Fig. 8).

The mean frequency (proportional to SP/MA) was on the average below normal due to the predominant increase of instability (MA) which corresponds to reduced regulatory movements. This instability type of postural ataxia contrasts to the fast balancing type with high MF found in cerebellar ataxia due to anterior lobe lesion and is similar to the reduced balancing activity in the vestibulo-cerebellar syndrome (Mauritz et al., 1979). Sway angle histograms were normal. Probably due to the rather small number of individuals tested, no correlation between the age of the lesion and our instability parameters could be found.

## Comment

The parameters defined in this paper allow for a clear differentiation between cerebellar ataxia in anterior lobe lesions and vestibular postural ataxia, despite a considerable intra- and interindividual variance. Since this differential diagnosis in most cases may also be done by pure clinical examination, our method may mainly be useful for quantification, documentation, and for follow-up studies which so far had to rely on very vague subjective reports and clinical estimates. Among the parameters tested, mean amplitude, sway path, sway area, and quotients quantifying visual stabilization of posture were found to be most useful. Only a few parameters allow for a visualization of features that seem specific to the site of a lesion in the cerebellum such as the predominance of *a-p* sway in the sway direction histograms of patients with an anterior lobe lesion. We feel that the sway direction and sway position histograms are particularly promising because of this effect. Considerably more experience has to be collected in future studies, however.

The analysis of the Fourier power spectrum has not been treated in this paper since it is a well-established method. Fourier analysis is of particular importance in early diagnosis of cerebellar atrophies where a specific 3 Hz tremor may frequently be demonstrated (Silfverskiöld, 1968; Dichgans et al., 1976; Mauritz et al., 1979).

All sway parameters show a significant increase in patients with anterior lobe cerebellar cortical atrophy. Furthermore, they indicate a disproportionately high balancing activity (SP, MF), together with a relatively low instability (MA), in those patients as compared to the group with labyrinthine defects. The latter group shows rather large mean amplitudes of sway (MA), short sway paths (SP) and consequently, low mean frequencies. Romberg's sign, which is the increase of postural sway with eyes closed in terms of the recorded parameters, is invariably positive in normals and in patients with vestibular as well as cerebellar lesions. The 'Romberg' quotient, i.e., sway with eyes closed divided by sway with eyes open (Njiokiktjien and van Parys, 1976) does not discriminate among those groups.

Quantitative estimation of several parameters in the literature was almost exclusively done on patients with labyrinthine lesions (Kapteyn and de Wyt, 1972; Taguchi et al., 1978). Cerebellar lesions were only investigated in a cursory manner (Njiokiktjien and van Parys, 1976). It is difficult to compare the quantitative results in our study with those in the literature, however, because of the different measuring devices and the different modes of filtering and analogue computation.

The clinical significance of this method is limited by the rather large intra- and interindividual variance of the sway measurements. We assume that the significance may be improved considerably by processing longer measuring intervals which should be possible by digital on-line analysis.

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