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Changes in Postural Muscle Dynamics as a Function of Age

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Histologic studies have demonstrated both a decrease in size and loss in number of type II muscle fibers with increasing age. Although these age-related histologic changes are believed to result in decreased strength and functional capacity, age-related changes in muscle force dynamics have not been clearly elucidated. Using vibromyographic (VMG) techniques, we recorded muscle activity of the soleus in 40 healthy adult volunteers spanning the age range of 20–82 years to test whether changes in postural muscle dynamics, in the frequency range of 0.1–50 Hz, were also associated with age. Although muscle dynamics below 15 Hz do not change with aging, the 30–50 Hz frequency components of the VMG were found to change significantly with advancing age ($r = -.619$, $p = .0001$). This was observed in both sexes independently. The observed age-related changes in muscle force dynamics demonstrate distinct physiologic alterations in muscle fiber activity. Further research will be required to fully elucidate the relationship between age-related changes in muscle fiber activity and other age-related conditions such as postural instability and osteoporosis.

ADVANCING adult age is responsible for a number of physiologic changes in the human body, including decreased aerobic and functional capacities. However, of all the age-related physiologic changes, loss of muscle strength has the most profound impact on the increasing number of elderly individuals (1). Reduction in muscle strength results in a greater tendency to fall, increased risk of hip fracture, impaired mobility, impairments in activities of daily living, and ultimately, results in increased levels of disability in older adults (2). This age-related loss of muscle strength can be attributed mainly to the loss of muscle mass, a phenomenon known as senile sarcopenia (3). Senile sarcopenia appears to be characterized histologically by selective atrophy of type II muscle fibers and overall decrease in muscle fiber number and cross-sectional area as a result of fatty and connective tissue infiltration (4). Indeed, using ultrasonography, Young and colleagues (5,6) found 25–35% reductions in cross-sectional area of the quadriceps muscle in older individuals compared to younger individuals. Using cross-sections of whole human vastus lateralis muscle, Lexell (7) found reductions in muscle cross-sectional area to commence in the third decade of life which were specifically due to decreases in type II muscle fiber size, whereas type I muscle fiber size remained unaffected.

Although age-related changes in muscle strength, composition, and structure have been well characterized, age-related changes in muscle force dynamics have not been adequately studied. While W. H. Wollaston (8) recognized in the 19th century that sustained muscular contraction generated vibrations with dominant frequencies between 14 and 36 Hz, recent technical advances have engendered further study and interest in muscle vibration. A new technique for measuring muscle vibration, known as vibromyography (VMG), records the muscle body accelerations generated during muscle contraction. Numerous investigators have reported a variety of mean and peak frequencies of VMG recordings, but there is no single stable frequency in a VMG because the signal is dependent on the muscle fiber composition of a particular muscle, the joint angle, and the level of contraction (9). Peak VMG signals are consistently found to occur within the range of 10–50 Hz and are believed to represent unfused motor unit recruitment pattern or activation rate (10). The amplitude of the VMG is dependent on the amplitude of fluctuations in muscle tension (11) and increases linearly with the level of contraction (12). Although electromyography (EMG) and VMG are both indices of muscle activation, their relationship to muscle force production are fundamentally different. EMG provides an indirect indicator of muscle force production by measuring the level of electrical activation of the muscle, whereas VMG measures the mechanical displacement of the muscle which reflects muscle fiber contraction. VMG is a more accurate measure of muscle force production and more directly reflects muscle contractile properties than the EMG (13). Thus, VMG represents an ideal tool for noninvasively measuring dynamic force output of muscle contraction.

The goal of this study, therefore, was to utilize VMG techniques to test the hypothesis that there are significant changes in muscle force dynamics associated with aging. We measured the dynamics of postural muscle activity during quiet standing in a study population spanning 20–82 years in age.

Materials and Methods

Subjects

The study was composed of 40 healthy, nonpregnant adult volunteers recruited from the Health Sciences Center at Stony Brook and from the local community. For inclusion in the study, volunteers were required to have no current or past history of connective tissue disorders, neuromuscular disorders, osteoporosis, scoliosis, alcohol abuse, or prolonged steroid use. The volunteers were excluded if they participated in competitive or collegiate athletics or weight-lifting programs. All volunteers were ambulatory and independent in activities of daily living. The research protocol was approved by the Committee on Research Involving Human Subjects at the State University of New York at Stony Brook. Written consent was obtained for all volunteers.

Measurements

Weight and height were measured for each subject. The body mass index (BMI; weight/height^2) was calculated for each volunteer. VMG signals were recorded using a low mass accelerometer (EGAX-F-25, Entran Devices, Inc., Fairfield, NJ) weighing 0.5 g with a range of ±25 g and sensitivity of 4.16 mV/g. The VMG signal was amplified (gain = 1000) using a bridge amplifier (Model 2120, Vishay Instruments, Measurements Group, Raleigh, NC) and output from the bridge amplifier was passed through a differential amplifier (Model AK-4LN, MetaMetrics, Inc., Carlisle, MA; gain = 50) with low-pass filtering (120 Hz) to prevent aliasing. Spectral analyses were obtained over the frequency range of 0.1–50 Hz at a sweep rate of 1 Hz per 3 seconds, using an analog low-frequency spectrum analyzer (Quantech, Quan-Tech Laboratories, Inc., Whippany, NJ). Output from the spectrum analyzer was digitized at 34 Hz.

Protocol

For inclusion in the study, all volunteers were required to respond to a questionnaire regarding their general health, past medical history, physical condition, exercise, normal daily levels of physical activity, and occupation. The questionnaire was used solely as an exclusion device. After completion of the questionnaire, volunteers removed their shoes and stockings for the duration of the experiment and anthropomorphic measurements were taken.

VMG signals were obtained from the soleus because it is the only muscle of the triceps surae that is active during quiet standing, as shown by EMG studies, and is the most superficial of all the muscles active during posture (14). The lower leg was palpated at the inferior edge of the gastrocnemius and the accelerometer was attached just inferior to the edge of the muscle belly of the gastrocnemius on the posterior aspect along the midline of the leg. Five recordings of the right soleus were made during a period of quiet sitting and five recordings made during a period of quiet standing. Recordings began after a 10-second period of quiet sitting or standing and lasted a total of 147 seconds each. This protocol was repeated for the left soleus for a total of 20 recordings per subject. During the passive phase of the protocol, the subject sat on a chair with a backrest. The height of the chair was adjusted to keep the knees and hips at 90-degree flexion and the feet flat on the floor. During the active phase of the protocol, the subject was asked to stand quietly, not to shift their weight from side-to-side or forward-backward, and to keep their arms at their side. No contact with any other objects or structures were permitted during the recording. Subjects were encouraged to rest in between recordings if they felt tired from standing.

Data Analysis

The average spectrum was calculated from five right and five left soleus spectral recordings using a data analysis software package (DAN, Triakis, Inc., Los Alamos, NM) for both standing and sitting such that each standing or sitting spectra was the average of 10 recordings. Subtraction spectra of active (standing) minus passive (sitting) were calculated for each volunteer. The human soleus is composed of 20% fast-oxidative-glycolytic (FOG) or type II muscle fibers and 80% slow-oxidative (SO) or type I muscle fibers (15). Because human SO muscle fibers are reported to have mean firing rates of 11 Hz [range 5–20 Hz; (16)] and FOG muscle fibers have mean firing rates of 31 Hz [range 15–50 Hz; (17)], we focused on two particular frequency bands of the VMG spectra: 0.1–15 Hz and 30–50 Hz. The separation of the VMG spectra into these distinct frequency bands allows the specific detection of primarily SO fiber activity within the 0.1–15 Hz frequency band and primarily FOG fiber activity within the 30–50 Hz frequency band of the VMG spectrum. To quantify the recordings, each spectrum was integrated over the ranges specified as well as over the total spectrum (0.1–50 Hz). The integrated values for each frequency range were normalized by their respective bandwidths.

The association of the spectral content in the entire spectrum and the two specific frequency ranges with age were evaluated by linear regression (StatView, Abacus Concepts, Inc., Berkeley, CA). In addition, a forward stepwise elimination model of multiple regression was used to determine which variables in combination were most predictive of spectral content. A critical F ratio of 5.343, corresponding to the 95th quantile ($p = .05$), was used for 1 and 39 degrees of freedom to determine independent variables that were entered or removed from the multiple regression. The independent variables considered were age, weight, height, and BMI (weight/height^2). In order to determine the effect of gender, a forward stepwise elimination model of multiple regression was repeated with gender entered as an independent variable. The correlations of the three frequency ranges (0.1–15 Hz, 15–30 Hz, and 30–50 Hz) with respect to age were compared using standard least-squares multiple regression to determine whether the correlations are significantly different from each other (JMP, SAS Institute, Inc., Cary, NC). Unpaired two-group t tests were used to compare male and female subjects. p values at levels of .05 or less were considered statistically significant.

Results

The volunteers ranged in age from 20 to 82 years with a mean age of 43.2 ± 18 years (mean ± *SD*). Weight ranged from 46.4 to 94.5 kg with a mean weight of 69.1 ± 12.5 kg. Height ranged from 152.4 to 188.0 cm with a mean height of 168.2 ± 9.3 cm. BMI ranged from 18.7 to 36.4 kg/m^2 with a mean BMI of 24.7 ± 4.2 kg/m^2. Males were significantly taller and heavier, whereas females were significantly older (Table 1).

Table 1. Physical Characteristics

	Total population ($n = 40$)	Male subjects ($n = 18$)	Female subjects ($n = 22$)
Age (years)	43.2 ± 18	35.9 ± 16*	49.1 ± 18*
Weight (Kg)	69.1 ± 12.5	75.9 ± 10.7§	63.5 ± 11.3§
Height (cm)	168.2 ± 9.3	175.23 ± 7.4†	162.36 ± 6.3†
BMI (kg/m^2)	24.7 ± 4.2	25.3 ± 3.6	24.1 ± 4.6

Note: BMI = body mass index. All numbers are mean ± *SD*.
$^*p < .05$; $^\dagger p < .01$; $^\S p = .001$.

Typical VMG recordings (Figure 1) from a 23-three-year-old volunteer illustrates the differences in VMG recording between active muscle during standing and the passive resting state during sitting. The sitting data consist primarily of room vibration and system noise with magnitude inversely proportional to frequency. Subtraction spectra, obtained by subtracting the VMG signal of the passive recording from the VMG signal of the active recording, shows the contribution of muscle activity to the VMG signal (Figure 2). Reproducibility of the VMG recordings was determined by repeated measures in a population of six individuals (three men, three women, ranging from 24 to 56 years of age) over a period of 4 days. Average coefficient of variation for this series of measurements was 27% (31% for the female group, 24% for the male).

The influence of age and gender on the subtracted spectra can be seen in typical recordings obtained from a 27- and a 73-three-year old woman (Figure 2a) and from a 23- and a 75-year-old man (Figure 2b). The subtraction spectra of the young volunteers are dominated by frequency components greater than 25 Hz. Conversely, a substantially attenuated signal is seen above 10 Hz in the older volunteers, with an increase in activity evident below 2 Hz.

The correlation matrix (Table 2) summarizes the statistics and relationships between VMG frequency content, anthropomorphic measurements, and age of all subjects. Unless otherwise noted, both male and female subjects were included in the analysis. The total integrate under the VMG curve (0.1–50 Hz) is positively correlated with height ($r = .425$, $p = .0063$) and negatively correlated with age ($r = -.484$, $p = .0015$; Figure 3). Although the integrated 0.1–15 Hz frequency components show little correlation with age or height, the integrated 30–50 Hz frequency component demonstrates a significant positive correlation with height ($r = -.449$, $p = .0037$) and negative correlation with age ($r = -.619$, $p = .0001$; Figure 4).

Forward stepwise elimination model of multiple regression with gender excluded as an independent variable reveals that age alone is predictive of both the total integrated VMG and the integrated 30-50 Hz frequency component of the VMG. Of the independent variables considered in the multiple regression, only age exceeded the critical F value of 5.343 ($p = .05$) with an F to remove = 23.666 ($p < .00001$) and 11.637 ($p = .0015$) for 30–50 Hz and 0.1–50 Hz, respectively. Other independent variables considered in the stepwise regression did not achieve the critical F value and did not add significantly to VMG predictability. The partial correlation coefficient for height was .22 and .25 with an F to enter of 1.876 (*NS*) and 2.469 (*NS*) for 30–50 Hz and 0.1–50 Hz, respectively.

When forward stepwise elimination model of multiple regression was performed with gender added as an independent variable, gender and age were the only two variables to remain in the regression equation. Again, none of the other independent variables considered in the stepwise regression achieved the critical F value. The F to remove for age was 5.613 ($p = .02$) and 15.034 ($p = .0004$) for 0.1–50 Hz and 30–50 Hz, respectively. The F to remove for gender was 9.029 ($p < .01$) and 6.973 ($p = .01$), for 0.1–50 Hz and 30–50 Hz, respectively.

Least-squares multiple regression was used to assess the independent association of each frequency range (0.1–15 Hz, 15–30 Hz, and 30–50 Hz) with age. This analysis revealed that

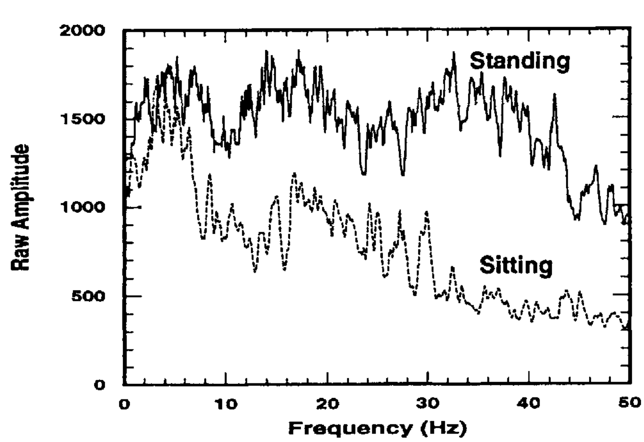

Figure 1. Average VMG spectra during quiet standing (active) and quiet sitting (passive) recorded from a 23-year-old man.

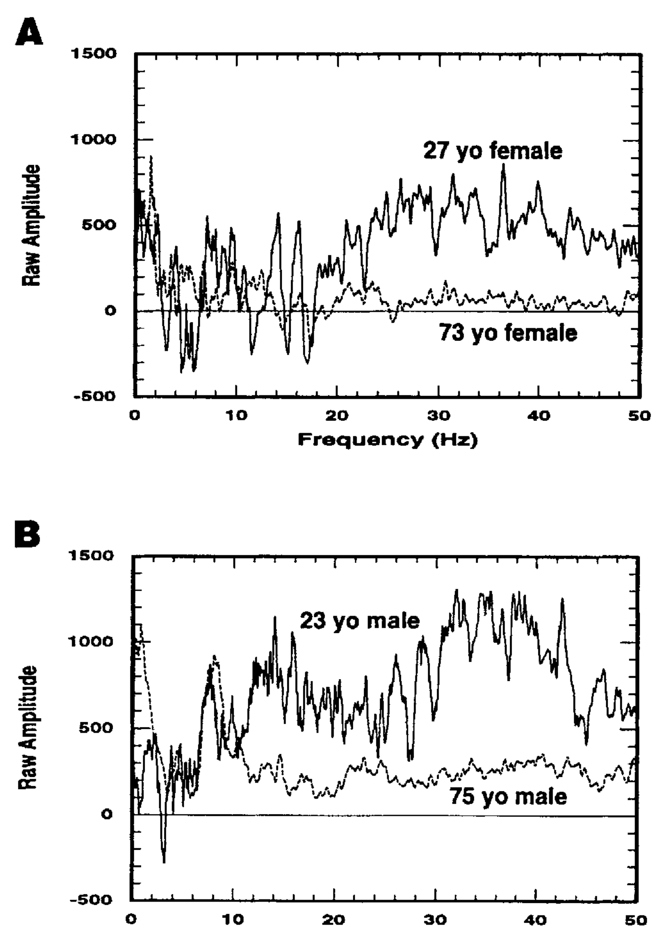

Figure 2. Muscle dynamics in *(a)* a 27-year-old woman (*solid*) and a 73-year-old woman (*dashed*) and *(b)* a 23-year-old man (*solid*) and a 75-year-old man (*dashed*) as assayed by the subtraction spectra. Subtraction spectra are calculated by subtracting passive VMG from active VMG and allow comparisons among the subjects. The subtraction eliminates the system noise, room vibrations, and other confounding vibrations.

Table 2. Correlation Matrix

	Age	BMI	Weight	Height	0.1–15 Hz	15–30 Hz	30–50 Hz
BMI	.341*	1					
Weight	.022	.799***	1				
Height	–.481**	–.143	.475**	1			
0.1–15 Hz	–.143	–.003	.164	.269	1		
15–30 Hz	–.306	.046	.210	.279	–.485**	1	
30–50 Hz	–.619***	–.074	.215	.449**	.325*	.662***	1
0.1–50 Hz	–.484**	.031	.293	.425**	.612***	.888***	.877***

Note. BMI = body mass index.
$^*p < .05$; $^{**}p < .01$; $^{***}p = .0001$.

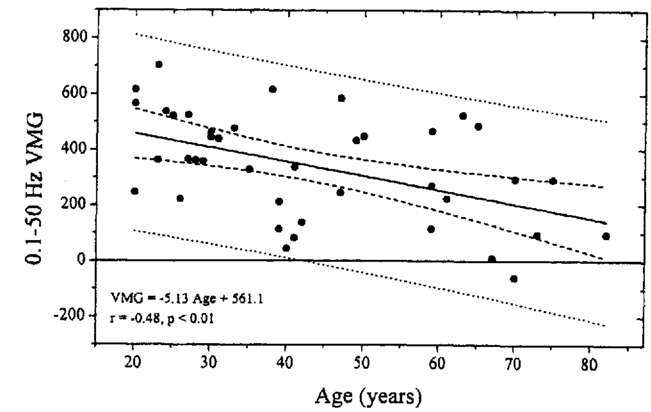

Figure 3. Relationship between total integrated VMG (normalized 0.1–50 Hz frequency content) and age. The linear regression best fit (*solid line*) is shown. The 95% confidence band for the regression line lies between dashed lines and the 95% prediction band for the regression line lies between dotted lines.

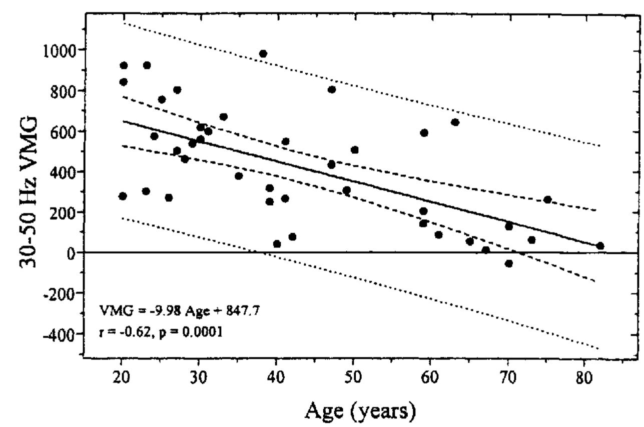

Figure 4. Relationship between integrated 30–50 Hz VMG (normalized 30–50 Hz frequency content) and age. The linear regression best fit (*solid line*) is shown. The 95% confidence band for the regression line lies between dashed lines and the 95% prediction band for the regression line lies between dotted lines.

the 30–50 Hz frequency component accounts for the majority of the changes observed with age with an F ratio equal to 18.618 ($p = .0001$). The F ratios for 0.1-15 Hz and 15-30 Hz were 0.0055 (*NS*) and 0.9478 (*NS*), respectively.

In addition, the difference in integrated VMG between male and female subjects was analyzed using two-group unpaired t tests (Table 3). Total integrated VMG in male subjects was significantly higher than total integrated VMG in female subjects. When the VMG signal was analyzed by frequency bands, the male subjects showed a significantly higher integrated 30–50 Hz frequency component than the female subjects. However, the significant correlation between the integrated 30–50 Hz frequency component and age in the total population holds true for both male and female subpopulations (males: $r = -.636$, $p = .004$, females: $r = -.458$, $p = .032$).

Table 3. Average Integrated VMG

	Total population (n = 40)	Male subjects (n = 18)	Female subjects (n = 22)
0.1–15 Hz	268.93 ± 141.01	310.56 ± 118.88	234.87 ± 150.95
15–30 Hz	292.74 ± 223.79	387.44 ± 196.08	215.257 ± 218.87
30–50 Hz	416.88 ± 289.35	581.25 ± 280.84*	282.39 ± 222.56*
0.1–50 Hz	339.70 ± 190.24	451.77 ± 154.15*	248.00 ± 168.52*

Note. VMG expressed as unitless numbers. All numbers are mean ± *SD*.
$^*p < .001$.

Discussion

In this study, we use modern VMG techniques to determine whether the dynamics of a specific postural muscle change with increasing age. We show that a significant decrease in the soleus VMG signal does occur with increasing age. Specifically, the 30–50 Hz frequency component of the VMG signal is seen to decline with age at a rate of 1.2% per year. Moreover, this component is identified as the main component of the soleus VMG signal that varies as a function of age, as we find no significant correlation between the 0.1–15 Hz frequency components and age. The strong negative correlation between the 30–50 Hz frequency component and age is consistent with loss of muscle strength secondary to reductions in the absolute number and decrease in size of type II muscle fibers commencing during the third decade of life (3,4).

The decrease in soleus muscle activity is consistent also with the observation that elderly people have lower postural stability than younger individuals. Elderly persons have a greater tendency to fall, greater impairment in mobility, and greater difficulty with activities of daily living (2). These deficits in posture, balance, and gait are secondary to the decline in muscle mass and strength. For example, Whipple and colleagues (18) demonstrated that ankle weakness, particularly involving the plantarflexors (i.e., triceps surae), was an important underlying factor in poor balance. More recent work has confirmed that muscle strength is a major factor in the maintenance of standing posture (1). Because selective type II muscle fiber atrophy has been determined as the main cause of age-related decline in muscle strength (3), the loss of the 30–50 Hz component of the VMG signal in the elderly persons may reflect the loss in dynamics contributed by type II fiber activity in the soleus.

The correlation between the total integrated VMG signal and the integrated 30–50 Hz frequency component with age was significant for both male and female subpopulations. This indicates that the same physiologic muscle changes occur in both men and women as a function of age. However, male subjects did have a significantly greater total VMG and 30–50 Hz frequency component signals than female subjects, consistent with the fact that men have greater muscle mass per unit skeleton than women (19,20). Although significantly lower levels of VMG activity were observed in the female subjects, the female subjects were also significantly older, and there is recent evidence to suggest that women undergo an accelerated sarcopenia in the perimenopausal period (21). Due to the design of the study, the possibility that the differences between men and women can be attributable to differences in age alone cannot be excluded.

Modern VMG techniques involve the use of highly sensitive and accurate accelerometers, so there are a number of factors that can affect the VMG recording. The most important factor involves the use of accelerometers. Accelerometers differ from alternative acoustic recording devices (i.e., hydrophones, electric condenser microphones, cardiac stethoscopes) in that vibrational accelerations are recorded rather than displacement or velocity. However, the use of accelerometers provides protection from confounding technical factors such as changes in contact pressure, low signal-to-noise ratio, and vibrational interface properties that can occur with other recording devices. In addition, signal-to-noise ratios are improved at higher frequencies where displacements are small, because the recorded acceleration will be proportional to the square of the frequency.

An important factor associated with this VMG study is that it does not rely on the use of maximal voluntary contractions. Human muscle studies commonly express muscle activity in terms of maximal voluntary contraction which is affected by effort and motivation and may not be reflective of the muscle's true maximum force (22). In this study, subject motivation is not a factor because little or no conscious effort is required to sit or stand quietly in normal individuals and the position of each posture (sitting or standing) was kept constant during different recordings from the same subject and among all subjects.

This study demonstrates that one aspect of the aging process is a dramatic change in the high frequency dynamics of postural muscle contraction. The high frequency (30–50 Hz) components of muscle activity decrease significantly with increasing age, an effect which can be expected to result in decreased higher frequency force components in the skeletal system. Loss of the dynamics of muscle contraction in elderly persons therefore represents a reduction in the mechanical stimuli which may be most critical for bone maintenance, because bone tissue appears to be more sensitive to higher frequency loading (>10 Hz) for maintenance of bone mass and stimulation of new bone formation (23). This degeneration of the high frequency skeletal strain components with aging occurs in both males and females and may explain the difficulty in maintaining bone mass during aging despite aggressive physical and/or pharmacological interventions. VMG analysis appears to provide a simple, direct, noninvasive means to characterize age-related sarcopenia in humans which may prove to be important in a variety of clinical applications.

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