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ORIGINAL ARTICLE

S. Okada · K. Hirakawa · Y. Takada · H. Kinoshita

Relationship between fear of falling and balancing ability during abrupt deceleration in aged women having similar habitual physical activities

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Abstract The purpose of this study was to evaluate the influence of fear of falling on balancing ability during abrupt deceleration in aged women. The subjects were 20 women aged between 67 and 75 years. They were classified into two groups, one having a fear of falling (group FF, $n = 10$) and another without this fear (group NFF, $n = 10$). The two groups had similar daily physical activities. Changes in the centre of foot pressure (CFP) were measured during postural sway following horizontal deceleration of the force platform on which they were standing, and the response time and CFP displacement were evaluated. In addition, the electromyogram (EMG) onset in the tibialis anterior muscle and medial gastrocnemius muscle during abrupt deceleration, and its difference between the two muscles were measured, and the relative level of co-contraction of antagonistic muscles (the co-contraction index, CCI) in the lower extremity muscle group was calculated. The response time and CFP displacement immediately after abrupt deceleration were significantly higher in group FF than in NFF ($P < 0.05$). The EMG onset in the two muscles did not significantly differ between the two groups. The difference in EMG onset between the two muscles was significantly lower in group FF than in NFF ($P < 0.05$). The CCI was significantly higher in group FF than in NFF ($P < 0.05$). These results suggest that there were negative effects of a fear of falling on the balancing ability immediately after abrupt deceleration. This may be because a fear of falling increases the co-contraction of antagonist muscles in the lower extremity muscle group.

Keywords Fear of falling · Balancing ability · Abrupt deceleration · Aged women · Co-contraction index

S. Okada (✉) · K. Hirakawa · Y. Takada
Faculty of Human Development, Kobe University,
3–11 Tsurukabuto, Nada-Ku, Kobe 657–8501, Japan
E-mail: shuokada@kobe-u.ac.jp
Tel.: +81-78-8037814
Fax: +81-78-8037929

H. Kinoshita
School of Health and Sports Sciences,
University of Osaka, 1–17 Machikaneyama-cho,
Toyonaka, Osaka 560–0043, Japan

Introduction

Falling is one of the most serious problems associated with aging among the human population (Maki et al. 1991; Perry 1985). Falling has been found to be a complex, multifactorial problem, and deteriorated postural control has often been cited as a major contributing factor (Barlet et al. 1986; Berg 1989; Lord et al. 1992; Nevitt et al. 1991). Postural control has been reported to require a complex interaction of musculoskeletal and neural systems (Shumway-Cook and Woollacott 1995).

In addition to this physiological factor contributing to postural control, Spirduso (1995) suggested a fear of falling as a psychological factor contributing to the lack of maintenance of balance. McAuley et al. (1997), Tinetti et al. (1990), and Walker and Howland (1990) reported that about 50% of aged people with a fear of falling have experienced falling. Maki et al. (1994), who carried out a 1 year follow-up survey, reported that aged people who had a fear of falling tend to fall more often than those not having this fear. These studies suggest an influence of a fear of falling on postural stability in the standing position.

On the other hand, there has been only one study, by Maki et al. (1991), on the relationship between a fear of falling and the ability to maintain balance following a perturbation of an upright stance. They analysed the centre of foot pressure (CFP) during spontaneous-sway and induced-sway tests and found a significantly poorer balancing ability in elderly people having a fear of falling in spontaneous-sway tests but no significant association between a fear of falling and CFP displacement in induced-sway tests using a movable platform. Based on these results, they stated that they could not ascertain whether a fear of falling affected balance-test performance in an artefactual manner, or whether the fear and the poorer performance were related to a true deterioration in postural control.

Thus, since no consistent conclusion has been obtained on the influence of a fear of falling on the balancing ability during perturbed stance, the necessity for more experimental studies on the relationship between a fear of falling and postural control has been recognised (Shumway-Cook and Woollacott 1995).

Vellas et al. (1987), Shumway-Cook and Woollacott (1995), and Spirduso (1995) have suggested that self-imposed restrictions on habitual physical activity due to a fear of falling induce a deterioration in balance. Inactivity has been reported to reduce muscle strength, joint flexibility, and other neuromotor functions, resulting in a deterioration in balance (Vellas et al. 1987; Spirduso 1995).

This indicates the influence of a fear of falling and habitual physical activity as psycho-physical factors influencing balancing ability. Therefore, the influence of habitual physical activity should be taken into account when the possible influences of a fear of falling on balancing ability during a perturbation of stance are being evaluated. However, to our knowledge, there have been no studies that have evaluated the relationship between a fear of falling and the balancing ability during a perturbed stance after eliminating the influence of habitual physical activity.

To clarify the influence of a fear of falling on balancing ability during abrupt deceleration, we have compared CFP displacement after abrupt deceleration in aged women having similar physical activities, between those having a fear of falling and those not having this fear. In addition, muscle activity in the lower extremity muscles during abrupt deceleration, and motor ability, have been compared between the two groups, and the relationship between a fear of falling and balancing ability has been discussed.

Methods

Subjects

Of 132 healthy women aged 65 years or over, 46 with a marked or some fear of falling (fear of falling group, group FF) and 53 with no such fear (no fear of falling group, group NFF) were selected (total, 99 women) using the method of Maki et al. (1991). Next, 10 women were selected from each of the two groups so that the two were matched for age, body height and mass, and physical activity. Physical activity was assessed using the questionnaire for investigating the activities of daily living described by Takada et al. (1996), and habitual physical activity (a physical activity score) was estimated. Falling was defined according to the definition of falling produced by Tinetti et al. (1988), as follows: a stoppage of the body on the ground or at a lower site, but excluding falling due to endogenous causes or unavoidable accidents. Using the report by Shumway-Cook et al. (1997) as a reference, the number of falls during the past 6 months was investigated. As a result, it was found in group FF that 8 subjects had experienced falling once; in group NFF, 3 subjects had also experienced falling once. A χ-square analysis showed that the two groups did not have significantly different distributions of falling.

Table 1 shows the mean values and standard deviations of age, body height and mass, foot length and physical activity score in the two groups which showed no significant differences. Interviews with the subjects concerning their medical histories indicated that none of them had disorders of the neuro-muscular system or were receiving therapy with hypotensive drugs. After explanation of the purpose and safety of this study, consent was obtained from all subjects.

Table 1 Characteristics of the subjects

	Fear of falling		No fear of falling	
	Mean	SD	Mean	SD
Age (years)	70.9	2.4	69.5	2.6
Height (m)	1.535	0.027	1.540	0.029
Body mass (kg)	55.4	2.8	56.7	2.9
Foot length (m)	0.229	0.006	0.230	0.005
Physical activity score	11.2	1.8	12.2	1.7

Apparatus

The experiment setup consisted of a platform for standing on, an electromyogram (EMG) recorder, and a personal computer. The platform for standing on was built using a force plate (Patela Inc., model K40, Tokyo, Japan) which was supported in two sliding rails fixed to an iron plate. The force plate could be moved horizontally using a solenoid (Meikosha Inc., model S12, Tokyo, Japan) to introduce an abrupt accelerated postural perturbation in a posterior direction. The plate was stopped after moving a predetermined distance by a buffer to generate a sudden deceleration. In the present study, the platform moved at 182 (± 8) $mm{\cdot}s^{-1}$ in an anterior direction. An anterior acceleration of the plate (mean, about 0.6 *g*; 70 ms) was initially applied, followed by a posterior deceleration (about –5.1 *g*, 10 ms).The distance moved was set to be 15 mm, which was the maximal distance that would not cause falls in the present elderly subjects.

Surface EMG was obtained from the tibialis anterior muscle and the medial head of the gastrocnemius of the right leg. Bipolar electrodes (10 mm diameter silver discs) were placed on the skin around the probable motor point of the muscles (Zipp 1982) at a longitudinal separation of 20 mm. The EMG signals were amplified using bio-amplifiers (Nippon Denki Sanei Inc, model 511, Tokyo, Japan) and stored in a personal computer via an A-D converter sampling at 1,000 Hz. The time constant of the amplifier was set at 0.03 s. Care was taken that the inter-electrode resistance was 10 kΩ or less (Fig. 1).

Procedure

The subjects with their feet bare placed the posterior surfaces of the two heels 0.2 m in front of the posterior edge of the force plate. They assumed a standing position at rest with the medial surfaces of their feet apart and parallel (distance, 0.1 m) and with both hands loosely attached to the sides of their bodies. The CFP in their maximal forward and backward leaning positions were then measured for each subject. During the experiment, the subjects were instructed to keep their CFP at the mean location of these two maximal CFP. The mean CFP in the range of motion of CFP in the anteroposterior direction was 0.103 (SD 0.01) m forwards from the heel in the subjects. The subjects then looked at a target placed 2 m before them at eye level, and the postural disturbances were then applied in this state. To minimize the subjects' anticipation of platform movement time, the delay between an auditory cue or ready call to the platform movement was altered within 1 to 3 s in a random manner. In our pilot study, no training effects were observed in this experimental task until the third measurement, which agreed with the findings of some previous studies on postural control (Alexander et al. 1992; Studenski et al. 1991). The mean

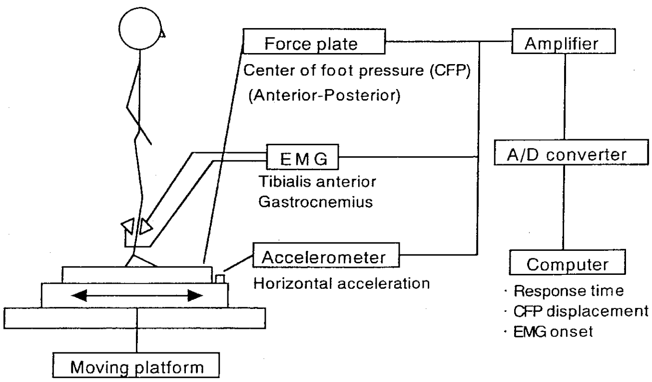

Fig. 1 A diagram of the experimental apparatus for providing an unexpected perturbation by an abrupt deceleration. *EMG* Electromyogram

value of three measurements was used as the representative value for each condition in each subject. The amplified signals from accelerometers on the platform and the EMG were A-D converted at sampling rate of 1,000 Hz. All of these data were stored in a personal computer for off-line analysis.

Analysis

Data between 2 s before and 3 s after the onset of movement of the plate were subjected to the analysis of postural control.

As shown in Fig. 2, response time variables (T1, T2, T3) and displacement variables (D1, D2, D3, and D4) were measured based on anteroposterior displacements of CFP. The time from the initiation of plate movement to the D1 point was similar among the subjects [122.9 (SD 6.8) ms].

For EMG data from the tibialis anterior and gastrocnemius muscles, the time from the initiation of plate movement to the first appearance of muscle activity (muscle activity latent time: EMG onset) was measured. The appearance of the EMG activity was determined when the signals exceeded the 2 SD above or below the mean of first 50 ms data in the 1-s period before plate movement (Stelmach et al. 1990). This EMG onset was determined firstly using a computer algorithm and secondly checked using an interactive graphics procedure. In addition, the difference in EMG onset between the tibialis anterior and gastrocnemius muscles was obtained.

To evaluate the relative level of co-contraction of the tibialis anterior and gastrocnemius muscles observed immediately after abrupt deceleration, the co-contraction index (CCI) was calculated using the method of Falconer and Winter (1985) and Maki and Ostrovski (1993). The CCI was calculated over the 200 ms interval beginning at the onset of activity. How the CCI was calculated from the scaled EMG signals is illustrated in Fig. 3.

The mean value in three measurements was calculated and used as the representative.

Measurement of motor ability

The following items were measured: dorsal flexion and plantar flexion strengths, knee extension and knee flexion strengths, range of joint motion of dorsal flexion, plantar flexion, hip extension, and hip flexion, one leg balancing time with eyes open and closed, and CFP fluctuations in the standing position at rest. In addition, sit-to-stand and walking tests were performed. Dorsal flexion and plantar flexion strengths were measured using a muscle force measurement device for the lower extremities (Wami Inc., model LS-10, Tokyo, Japan). As knee extension and flexion strengths, the torques at angular velocities of 1.05 and 3.14 rad·s^{-1} were measured using a Cybex dynamometer (Lumex Inc., CYBEX II, New York, USA).

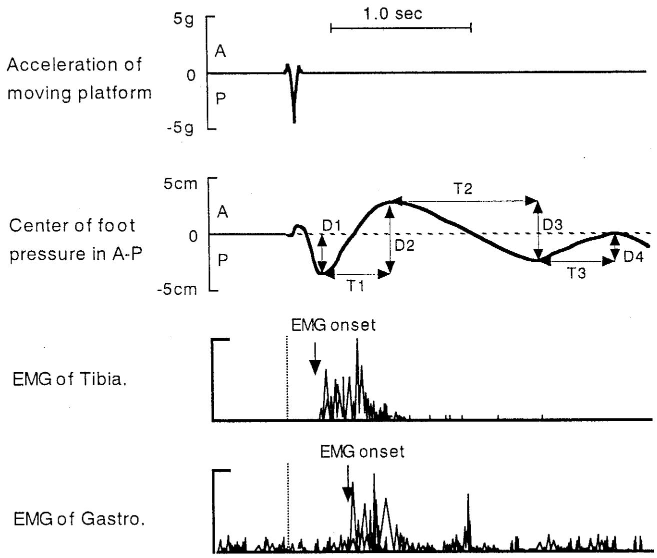

Fig. 2 Analysis of response time variables (*T1, T2, T3*) and CFP displacement variables (*D1, D2, D3, D4*) for centre of foot pressure data, and electromyogram (*EMG*) onset time in lower extremity muscles. *Tibia.* tibialis anterior, *Gastro.* gastrocnemius, *A* anterior, *P* posterior

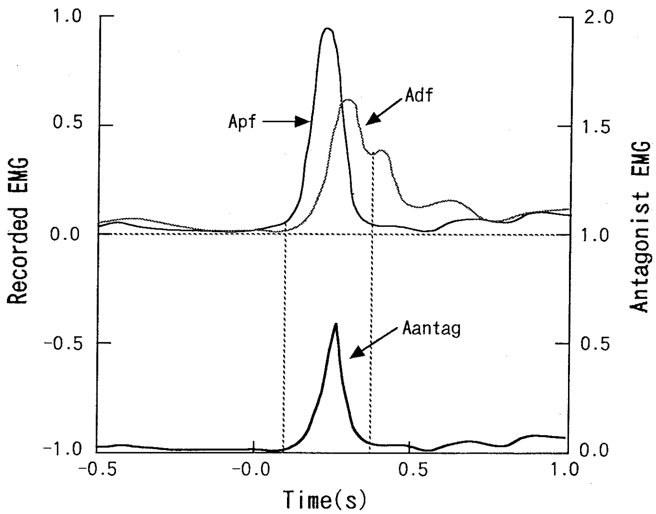

Fig. 3 Definition of the co-contraction index (CCI). Note, electromyogram (*EMG*) responses are scaled with respect to each other using results from isometric co-contraction trials before calculating the CCI. Plantarflexor EMG *thin line*, dorsiflexor EMG *wavy line*, antagonist EMG *thick line*, CCI = (2Aantag)/(Adf + Apf), where *Aantag*, *Adf* and *Apf* are the activities of antagonist, dorsiflexor and plantar flexor muscles, respectively

The range of ankle and hip joint motions was measured using a Todai type joint angle measuring apparatus (Yagami Co., model S-30, Tokyo, Japan). The CFP fluctuations were evaluated in terms of CFP displacement during a 20-s period with the eyes open and then closed using a Gravicorder (Anima Co., model GS-10, Tokyo, Japan). Sit-to-stand tests were performed by the method of MacRae et al. (1992), and the times required for standing-up from a chair for each of five times were measured. Walking tests were performed using the method of Himann et al. (1988) and Wolfson et al. (1990), and the walking speed and stride length during walking as fast as possible were measured or calculated.

Statistics

One-way analysis of variance (ANOVA) was used to determine the difference in the mean values of each parameter between the FF and NFF. A $P < 0.05$ was considered significant.

Results

Response time and CFP displacement in groups FF and NFF

The response time variables (T1, T2, T3) and CFP displacement variables (D1, D2, D3, D4) in groups FF and NFF are shown in Table 2. The response time variables and CFP displacement variables were higher in group FF than in group NFF. Significant differences were observed in T1 and D2 between the two groups ($P < 0.05$).

EMG onset in the tibialis anterior and gastrocnemius muscles and the difference in EMG onset between the two muscles in groups FF and NFF

The EMG onsets in the tibialis anterior and gastrocnemius muscles in groups FF and NFF are shown in Table 3. No significant difference was observed in the EMG onset of either muscle between the two groups.

The difference in EMG onset between the tibialis anterior and gastrocnemius muscles was 8.2 (SD 5.6) ms in group FF and 17.1 (SD 8.4) ms in group NFF, being significantly lower in group FF ($P < 0.05$). This suggests that the first appearance of EMG activity of the gastrocnemius muscle after the first appearance of EMG activity of the tibialis anterior muscle occurred significantly earlier in group FF than in group NFF.

CCI for the tibialis anterior and gastrocnemius muscles in groups FF and NFF

The CCI for the antagonistic muscles was 0.56 (SD 0.114) in group FF and 0.45 (SD 0.098) in group NFF, being significantly higher in group FF ($P < 0.05$).

Table 2 Response time variables (*T1, T2, T3*) and centre of foot pressure (*CFP*) displacement variables (*D1, D2, D3, D4*) of the fear of falling and no fear of falling groups

	Fear of falling		No fear of falling	
	Mean	SD	Mean	SD
T1 (ms)	541.4	92.2	439.9*	915
T2 (ms)	641.1	105.3	607.2	99.9
T3 (ms)	538.8	111.6	520.5	96.1
D1 (%)[a]	22.1	4.8	20.6	4.3
D2 (%)[a]	34.2	7.5	26.9*	6.0
D3 (%)[a]	27.7	7.7	23.2	5.9
D4 (%)[a]	10.7	4.2	8.1	2.8

[a]CPF displacements were normalized by dividing by foot length
*$P < 0.05$

Table 3 Electromyogram onset of tibialis anterior and gastrocnemius muscles

Muscle	Fear of falling		No fear of falling	
	Mean	SD	Mean	SD
Tibialis anterior (ms)	139.5	22.3	135.1	21.0
Gastrocnemius (ms)	147.7	27.6	152.2	25.1

Motor ability in groups FF and NFF

Table 4 shows dorsal flexion and plantar flexion strengths, knee extension and knee flexion strengths, range of joint motion of dorsal flexion, plantar flexion, hip extension, and hip flexion, one leg balancing time with eyes open and then closed, CFP displacement with the eyes open and then closed, sit-to-stand time, and walking speed and stride length in groups FF and NFF. Group FF showed a significantly shorter one leg balancing time with the eyes closed ($P < 0.05$) and a significantly longer CFP displacement with the eyes closed ($P < 0.05$) than group NFF. Excluding these items, no other significant differences were observed between the two groups.

Discussion

There were four response phases of CFP displacement during balance recovery after abrupt deceleration in this study (Fig. 2). The first response was a posterior CFP displacement due to the deceleration resulting from the moving plate hitting the buffer (passive phase: D1). Next, a response to this displacement occurred. In trying to recover CFP in orientation, but due to overshooting, the balance was destroyed, resulting in an anterior displacement (first active phase: T1, D2). A further recov-

Table 4 Motor ability assessments of the fear of falling and no fear of falling groups. *OLBT* One leg balancing time, *CFP* centre of foot pressure, CFP_{disp} CFP displacement

Variable	Fear of falling		No fear of falling	
	Mean	SD	Mean	SD
Dorsal flexion (N)	110.9	23.2	120.5	213
Plantar flexion (N)	200.1	61.7	230.0	62.2
Knee extention (N·m)	26.7	7.4	30.1	7.1
Knee flexion (N·m)	18.8	5.6	21.1	5.9
Dorsal flexion (°)	14.2	4.0	15.7	3.9
Plantar flexion (°)	52.4	7.5	55.5	7.4
Hip flexion (°)	72.8	8.9	75.2	8.8
Hip extension (°)	25.1	3.6	27.0	3.4
OLBT: eyes open (s)	30.7	18.8	38.2	19.1
OLBT: eyes closed (s)	3.4	7.7	11.3*	8.2
CFP_{disp}: eyes open (cm)	25.0	8.1	24.3	7.8
CFP_{disp}: eyes closed (cm)	43.7	10.3	33.8*	9.1
Sit-to-stand: 5 times (s)	7.7	2.8	7.0	2.2
Walking speed[a] ($m \cdot s^{-1}$)	1.82	0.52	1.99	0.33
Stride length[a] (m)	0. 63	0.11	0.69	0.09

[a]Walking as fast as possible
*$P < 0.05$

ery response to this displacement was observed, but overshooting occurred, and CFP was displaced posteriorly once again (second active phase: T2, D3). Subsequently, a response to recover CFP in orientation occurred (third active phase: T3, D4), gradually recovering the orientation of CFP.

Compared with group NFF, group FF showed a delayed response (T1) and marked CFP displacement (D2) in the first active phase. These results suggest that a fear of falling perhaps caused by the experience of a previous fall affects the recovery response immediately after abrupt deceleration but not the recovery responses in the second active and subsequent phases.

Concerning the mechanism of the influence of a fear of, or anxiety about, falling on balancing ability, Alexander (1994) and Maki et al. (1991) have proposed a hypothesis that aged people with these fears tend to stiffen their bodies when stance is perturbed. It has been suggested that this stiffening can be induced by co-contraction of antagonist muscles (Kearney and Hunter 1990). In this study, therefore, the CCI of the lower extremity muscles was calculated. Aged women with a fear of falling indeed had a higher CCI value, indicating a higher level of co-contraction in their lower extremity muscles than those lacking this fear.

Maki et al. (1991) suggested the following mechanism for co-contraction. Aged people with fear of a perturbation of their upright stance tend to adopt a stiffened strategy to minimize the risk of falling, and such compensation is achieved by stiffening of the postural control system by co-contraction of antagonist muscles. During the horizontal anterior-posterior movement of the moving platform, stiffening of the ankle due to co-contraction of the lower extremity muscles effectively acts on postural stability. However, during the abrupt decelerations due to the rapid movements and sudden stops of the plate in this study, an increase in the stiffness of the ankle may have resulted in direct transmission of the impact due to the sudden stop to the body centre of mass, increasing displacement. Considering the EMG onset time in the tibialis anterior muscle that was first activated, the influence of the co-contraction appeared in the first active phase. This may have resulted in the higher values in T1 and D2 in group FF with a higher co-contraction level than in group NFF.

Sato and Fujita (1998) have evaluated ankle joint viscoelasticity during changes in posture and vision and suggested a control of ankle joint viscoelasticity by a higher postural control system. Brooks (1986) described how muscle stiffness is controlled by both reflexes and upper levels of the central nervous system. These studies indicate that co-contraction of the lower extremity muscles is programmed from higher level neural systems. In this study there was a significantly earlier appearance of the first EMG activity of the gastrocnemius in group FF than in group NFF. Shumway-Cook and Woollacott (1995) reported modification of the postural control strategy based on cognition of abrupt acceleration in the aged. These findings seem to indicate that aged people with a fear of falling adopt a stiffened muscle strategy compared with those without this fear.

From the results of motor ability, no significant differences were observed in the muscle strength and flexibility of the legs and hips or the static balancing ability and mobility with the eyes open between groups FF and NFF. However, the response time and displacement after abrupt deceleration differed between the two groups, suggesting the influence of intrinsic factors other than motor ability on the balancing ability during abrupt deceleration. Maki et al. (1991) evaluated the possible relationship between a fear of falling and CFP displacement in the standing position at rest and observed a significant relationship between the two parameters when blindfolded but not with the eyes open. They suggested that this difference between the blindfolded and eyes-open states was due to a deterioration in proprioceptive or vestibular function in the fearful subjects. Based on this speculation, the significantly poorer results in terms of the displacement of CFP with the eyes closed in group FF than in group NFF in this study may also indicate deterioration in proprioceptive or vestibular function in group FF. In people with a fear of falling, deprivation of vision has been shown to enhance the influence of anxiety (Maki et al. 1991). Anxiety has been found to induce changes in γ-motoneuron activity (Ribot et al. 1986) and reduce attention (Stelmach et al. 1990), which may have caused the significant differences in the one leg balancing time and CFP displacement with the eyes closed between the two groups.

In conclusion, a fear of falling has negative effects on the balancing ability immediately after an abrupt deceleration. This may be associated with a more marked co-contraction of antagonist muscles in the lower extremity muscle groups in subjects with a fear of falling than those without this fear.

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