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# Mammosite Brachytherapy Dosimetry-Effect of Contrast and Air Interface on Skin Dose

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#### Abstract

High-dose-rate (HDR) brachytherapy is an effective internal radiation therapy procedure for treating malignant neoplasms. This technique is widely used in breast cancer treatments to destroy residual cancer cells surrounding the lumpectomy cavity following surgery. This is done by inserting a balloon catheter into the cavity that is inflated with saline as well as a medium of radiographic contrast. Then the radioactive isotope is positioned into the center of the balloon using an HDR unit to deliver the prescribed dose to a volume surrounding the balloon. Most of the currently available treatment planning systems (TPS) for brachytherapy, including Nucletron Oncentra, estimate dose using proprietary algorithms which use a pre-calculated dose metric derived from Ir-192 placed in a water phantom. However, they do not take into account variations in attenuation due to inhomogeneities in different tissues. This may lead to several questions: do these TPS estimate absorbed dose correctly within the target tissue? What are the effects on breast-air interface within the target volume? Does a radiographic contrast medium in the balloon alter the dose distribution calculated by TPS? These uncertainties and doses can be quantified by using the data recorded in a tissueequivalent patient phantom which is aided by a PN junction commercial diode detector and an electrometer. During this investigation, we used a cubical water phantom and Mammosite® single lumen balloon system to measure effects on breast-air interface, the diode detector was placed on the phantom wall to simulate the tissue air interface. Measured data were compared with predictions from the Oncentra TPS for the same geometry. These results may help quantify uncertainties in the predicted versus actual skin doses used during the treatments. This in turn could increase the clinicians' predictive power regarding potential excessive skin dose that could cause toxicity in patients.

**Keywords:** Brachytherapy; Radiation therapy; Brachytherapy dosimetry; Radial dose function; Breast cancer

# Introduction

Among many types of cancer, breast cancer causes the most deaths among women worldwide [1]. The United States alone reported 232,340 new breast cancer cases and 39,620 deaths due to breast cancer in 2013 [2]. It could be inferred using the rates from 2007-2009 that 12.38% of women born today might be diagnosed with breast cancer during their lifetime [3,4]. In Table 1, it shows increasing five-year relative survival rates in the United States [5]. The increasing rate of survival is thought to be due to the currently available advanced treatment options and early diagnosis.

There are mainly three types of cancer treatment techniques; radiation therapy, surgical procedures and chemotherapy. Radiation therapy is divided into two categories as external beam radiation therapy and brachytherapy. External beam radiation therapy (EBRT) is one of the oldest and effective techniques used to treat cancer. It delivers one or multiple beams of high-energy x-rays from an external source to a patient's tumor. During this technique, beams are generated outside the patient and are targeted at the tumor site [6]. The treatment time could vary from hours to several weeks depending on the type as well as the stage of the cancer.

Short distance treatment of cancer using radiation from a small encapsulated radionuclide source is called Brachytherapy [7]. It is performed by placing sources of radiation directly in or near the tumor to be treated. The dose of radiation has to be delivered continuously over a short period of time when temporary implants are used or over the lifetime of the source when permanent implants are used. The most common radiation modality used in brachytherapy is gamma photons.

There are two main types of brachytherapy treatments; intracavitary (where the sources are placed adjacent to the treatment volume, e.g., gynecology HDR treatments) and interstitial (where the sources are implanted inside the treatment volume, e.g., prostate LDR treatments) [7]. During this study, we focus on an intracavitary treatment method called MammoSite<sup>®</sup> brachytherapy.

Radiation may be used to treat breast cancer after the tumor has been removed with a lumpectomy. This technique is used to destroy residual cancer cells surrounding the lumpectomy cavity and it is known as accelerated partial breast irradiation (APBI) [1]. The device used to perform this procedure is called MammoSite\* catheter. It consists of a small silicon balloon connected to an external inflation lumen and an internal lumen for the passage of a high dose rate Ir-192 brachytherapy source [8]. It is placed into the lumpectomy cavity and inflated with a mixture of saline and radiographic contrast agent to fill the cavity [8]. The Ir-192 brachytherapy source is placed at the center of the balloon through a remote after loader to deliver dose to the treated volume.

#### Treatment

Treatment planning is the process of determining the target (tumor volume) and optimal technique for radiation delivery. After implantation of the MammoSite<sup>\*</sup> balloon catheter, the patient

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Year of diagnosis	Five year relative survival rate (%)
1975-1977	74.8
1978-1980	74.4
1981-1983	76.1
1984-1986	78.9
1987-1989	84.0
1990-1992	85.2
1993-1995	86.4
1996-1998	88.2
1999-2003	89.9
2004-2010	90.5

Table 1: Relative survival rates from breast cancer.

undergoes a CT scan. These CT imagers are used by the Nucletron treatment planning software to determine the area to be treated and those areas to be avoided. First, CT scans are transferred digitally to the treatment planning software to determine gross tumor volume (GTV), clinical tumor volume (CTV), and planning tumor volume (PTV). In Mammosite<sup>®</sup> brachytherapy, the excision of the lumpectomy cavity is defined as the gross tumor volume [9]. The PTV is defined as the breast tissue immediately surrounding the balloon to a measured 1 cm distance from the balloon surface [10].

In MammoSite\* technique, the PTV is considered to be the same as the CTV because of the implanted balloon moves with the target and, therefore, compensation for the variability of breathing motion is not needed. Contouring starts with definition of the GTV (also the planning target balloon surface) on a central slice of the CT scan. Then on each axial CT slice moving superiorly and then inferiorly [11]. Once all of the contours are drawn, then the CTV and prescribed dose limits are defined using iso dose surfaces.

Finally, the treatment planning software is used to design a source position sequence in various locations within the implant applicator to build a matching dose distribution according to the isodose surfaces. The amount of dwell time in one particular location determines the amount of dose delivered to that location (Figure 1).

#### Statement of the problem

Computer tomography (CT) simulation is used during the treatment planning procedure [11]. Therefore, the balloon is inflated with saline solution and contrast media so that it can be imaged through CT. However, the dosimetry of commercially available brachytherapy treatment planning systems such as Nucletron Oncentra is based on the assumption that the HDR brachytherapy source is located in a unit density water medium [12]. However, the balloon and its contents are no longer equivalent to water because of the contrast medium contains Iodine which has a high atomic number [8,13].

Most of the treatment planning systems for brachytherapy procedures including Nucletron Oncentra use water-based dosimetry. A homogeneous medium is assumed in this process. However, this may cause erroneous results because of the fact that the human body is heterogeneous. Therefore, these treatment planning systems assume that there exists a full scatter during the treatment regardless of the existing differences in tissue density in the human body. However, this assumption is not valid especially when the tissue beyond the prescription line is thin [14]. As an example, when the balloon is closer to breast-air interface, dose contribution due to backscatter radiation is smaller (Figure 2). Therefore, it is important that the treatment planning systems should consider patient anatomy and heterogeneity when it calculates the target dose rate. Otherwise, it will overestimate the dose rates in the areas of breast-air and breast-lung interfaces. Current radiation techniques do not have perfect efficacy. In addition, healthy cells are destroyed as well as malignant cells during the treatments [15]. This can result in significant patient morbidity. Delivering correct doses with an accuracy of +/-5% is important in order to successfully treat cancer with radiation [1]. Some treatment options have been created to spare healthy cells while still having enough radiation to damage cancerous cells [1]. However, if the cancer cells are not completely destroyed, then they could continue to grow and could cause the tumor to grow [15]. Therefore, it is very important to understand how radiation doses change due to contrast and breast air interface so that more accurate doses can be determined [15].

# Materials and Methods

#### Dose rate calculation

Most of the treatment planning systems used TG-43U1 model for the calculation of dose distributions around a brachytherapy source [16]. TG-43 is the first report on the dosimetry of sources used in interstitial brachytherapy [16,17]. Since then, the American Association of Physicists in Medicine (AAPM) introduced TG-43U1, an updated version of TG-43, because of the improved dosimetry methodologies and dosimetric characterization of particular source models [17].

Both of these reports introduce new parameters such as air kerma strength,  $(S_{\kappa})$  dose rate constant ( $\Lambda$ ) geometry function,  $(G_{L}(r,\theta))$  radial dose function,  $(g_{L}(r))$  and anisotropy function  $(F(r,\theta))$  to calculate brachytherapy dose [16,17]. These dosimetry parameters consider geometry, encapsulation and self-filtration of the source, the





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spatial distribution of radioactivity within the source and scattering in water surrounding the source [16,17]. According to this protocol, the absorbed dose rate distribution around a sealed brachytherapy source at point P (Figure 3) with polar coordinates (r,  $\theta$ ) can be determined using the following formalism [16,17]:

$$D(r,\theta) = S_k \cdot \Lambda \cdot \frac{G_L(r,\theta)}{G_L(r=1,\theta=\frac{\pi}{2})} \cdot g_L(r) \cdot F(r,\theta)$$
(1)

Air kerma strength,  $(S_{\kappa})$  represents brachytherapy source strength. It is defined as the product of air kerma rate at a calibration distance (d) in free space (which is usually chosen to be 1 m) along the perpendicular bisector of the source and the square of the distance  $(d^2)$  [16,17]. It is defined in the following equation.

$$S_k = K(d).d^2 \tag{2}$$

Where, K(d) is the air kerma rate. The unit of  $S_k$  is  $U(1U=1cGy. cm^2.h^{-1})$  [16].

The dose rate constant,  $\Lambda$ , for the source and surrounding medium is defined at  $P(r_{\theta}, \theta_{\theta})$  (One cm away from the source on its perpendicular bisector where,  $\theta = \frac{\Pi}{2}$ .  $\Lambda$  is expressed in units of *cGy*.  $hr^{-1}$ .  $U^{-1}$  [18] and it is defined as [16,17]:

$$\Lambda = \frac{D(r_0, \theta_0)}{S_t} \tag{3}$$

The value of the dose rate constant depends on the medium surrounding the radiation source since it indicates the rate of the energy which is absorbed by the medium [17,18]. Therefore, it is taking into account the factors such as source geometry, the spatial distribution of radioactivity within the source, encapsulation, and self-filtration within the source and scattering in the water surrounding the source [16,17]. The Geometry function,  $G_L(r,\theta)$  with units of cm<sup>-2</sup> describes the decrease in dose with distance from the source [16,18].

For point source [16,17],  

$$C(r, \theta) = r^2$$

$$G_{L}(r,\theta) = r^{2}$$
<sup>(4)</sup>

For line source [17],

$$G(r,\theta) = \frac{\theta_2 - \theta_1}{L.r.Sin(\)}$$
(5)

The radial dose function,  $g_L(r)$  takes into account absorption and scatters of radiation along the transverse axis of the source, and



normalized it to the value at 1 cm from the source.  $g_L(r)$  is determined from the depth dose measurements along the perpendicular axis of the source [17,18].

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$$g_{L}(r) = \frac{D(r,\theta).G_{L}(r,\theta_{0})}{D(r_{0},\theta_{0}).G_{L}(r,\theta)}$$
(6)

Anisotropy function,  $F(r,\theta)$  accounts for the angular variation of photon absorption and scattering in the medium and in the source encapsulation [16,17]. This function is expressed as a relative dose measurement, and it is normalized to the measurement at  $\dot{e}_{\theta} = \frac{\Pi}{2}$  for each value of r [16,17].

$$F(r,\theta) = \frac{D(r,\theta).G_L(r,\theta_0)}{D(r,\theta_0).G_L(r,\theta)}$$
(7)

**Dose calculation:** In calculating the absorbed dose value  $D(r,\theta)$ , it was assumed that the source strength was expressed as air kerma strength. The dose rate is obtained assuming that  $t_0 = 0$  [19]. If the dwell time of the <sup>192</sup>Ir source is T, then the total dose delivered at the point of interest is obtained using Eq. 6 [19]; where  $\lambda$  is the decay constant of the radio isotope.

$$D(r,\theta) = \int_{t=t_0}^{t=T} D(r,\theta,t_0) e^{-\lambda(t-t_0)} dt = D(r,\theta) \cdot \left(\frac{1}{\lambda}\right) \cdot (1-e^{-\lambda t})$$
(8)

But, mean life,  $\tau$  is equal to  $\frac{1}{\gamma}$ 

$$D(r,\theta) = D(r,\theta,t_0).\tau.\left(1 - e^{-\frac{T}{\tau}}\right)$$
(9)

When  $T \ll \tau$  exponential part of Eq. 9 becomes;

$$e^{\frac{T}{\tau}} \approx \left(1 - \frac{T}{\tau}\right) \tag{10}$$

Therefore, Eq. 9 can be expressed as

$$D(r,\theta) = D(r,\theta,t_0).\tau. \left[1 - \left(1 - \frac{T}{\tau}\right)\right]$$
(11)

$$D(r,\theta) = D(r,\theta,t_0).T$$
(12)

According to Eq. 12, it can be assumed that the dose,  $D(r,\theta)$ , at a particular point of interest,  $P(r,\theta)$ , is proportional to the dwell time, T. As a result, dwell time can be used as an indicator for the dose.

# Phantom

During this investigation, a clinical MammoSite\* balloon of 4 cm diameter was used. The MammoSite\* balloon was placed in the middle of the 30 cm x 30 cm x 30 cm water phantom. A 0.6 cc ion chamber was placed parallel to the catheter axis between the water surface and the balloon surface. The distance between the water surface and the middle of the chamber axis was set to 7 cm and the distance between the middle of the chamber surface to the MammoSite\* catheter axis was set to 5 cm. Both holders used during this investigation to hold the MammoSite\* catheter and ion chamber in place were made with tissue equivalent materials (similar electron density). These tissue equivalent materials and water simulate the attenuation and the scattering of soft tissues (Figures 4 and 5).

The phantom was filled with water and aligned with positioning







Figure 5: Schematic diagram of a 100 cGy isodose curve appear in treatment planning system.

lasers on the General Electric Lite Speed 4 slice Computed Tomography (CT) scanner. Then, the MammoSite<sup>\*</sup> balloon was then filled with 60 cm<sup>3</sup> of water, which representing 0% contrast. Later, helical CT scans were acquired throughout the phantom with a 1 mm slice width. Ten percent of water in the balloon was then replaced with an equal volume of a contrast solution and a repeat CT scan was acquired. The contrast medium used in this experiment is Iohexol which is sold under the trade name of Omnipaque 300 [20]. It has a molecular composition of  $C_{19}H_{26}I_3N_3O_9$  [20]. Omnipaque 300 contains 647 mg of Iohexol equivalent to 300 mg of organic iodine per milliliter [13]. The filling and scanning process of the MammoSite<sup>\*</sup> balloon catheter was repeated for the balloon filled with saline and 20% of radiographic contrast concentration.

# **Treatment planning**

The CT images of the phantom and the MammoSite\* balloon described previously were imported to the Nucletron Oncentra treatment planning system for contouring. After the contouring, a 100 cGy isodose curve was defined at 5 cm from the center of the balloon. The dose of 100 cGy was chosen for convenience and the whole process was repeated for all of the contrast concentrations.

# **Radiation delivery**

The treatment planning data from Nucletron Oncentra system was exported to the after loader. The phantom was placed on a stable counter top and the MammoSite<sup>®</sup> catheter was connected to the afterloader using a transfer tube. Treatment was delivered to the phantom. This was done twice to obtain an average measurement; both dwell times as well as ionization chamber readings were recorded. The same procedure was repeated for the rest of the contrast concentrations using the same MammoSite<sup>®</sup> balloon and phantom setup. The distance from the center axis of ion chamber to the center axis of MammoSite<sup>®</sup> catheter was kept at 5 cm.

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# **CT** simulation

The phantom was filled with water up to the center axis of ion chamber, where d is equal to zero. Then, helical CT scans were acquired throughout the phantom with 1 mm slice width using the same CT scanner described previously. The filling and scanning process was repeated for different d values, where d was equal to 1 cm, 2 cm, 5 cm and 7 cm. These different d values represent different tissue thicknesses beyond the point of measurements (Figure 6).

#### **Treatment planning**

The CT images of the phantom and the MammoSite\* balloon described previously were imported to the Nucletron Oncentra treatment planning system for contouring. As discussed in the Treatment Planning section, the 100 cGy isodose curve was defined at 5 cm from the center of the balloon. This was repeated for all of the over-lying water levels (Figure 7).

#### **Radiation delivery**

The treatment planning data from the Nucletron Oncentra treatment planning system was exported to the after-loader. As described in the Radiation Delivery section, the planned treatment was delivered to the phantom. This was done twice again to obtain an average measurement; both dwell times as well as ionization chamber





readings were recorded. The same procedure was repeated for the rest of the over-lying water thicknesses.

#### MammoSite® balloon catheter

The MammoSite<sup>®</sup> device used in this experiment consists of a singlelumen catheter and a silicone balloon (Figure 8). The external lumen was used as an inflation channel and the central lumen accommodates the radioisotope high (e.g., <sup>192</sup>Ir). The diameter of the spherical balloon was 4 cm when the fill volume equals to 60 cm<sup>3</sup>.

#### Ion chamber

The 0.6 cm<sup>3</sup> PTW Farmer ion chamber (Figure 9) that was used in this experiment is the most convenient as well as accurate dosimeter for measuring radiation dose in radiotherapy. Ionization chambers do not measure dose directly, thus it is necessary to convert measured ionization to absorbed dose by the chamber. This was done by multiplying the ion chamber reading with,  $N_{D,W}$  which was called the dose to water calibration coefficient. Before multiplying by,  $N_{D,W}$ , the raw readings from the ion chamber must be adjusted for environmental conditions as well as some aspects of operation of instrument as indicated in Equation 12.

$$M = M_{raw} P_{TP'} P_{ion} P_{pol'} P_{elec}$$
(13)

Where  $P_{TP}$  is a correction for the temperature (T) and pressure (P).  $P_{ion}$  is a correction factor for the incomplete collection of ions at electrodes in the ion chamber.  $P_{pol}$  is the polarity correction factor for differences in the sensitivity of the ionization chamber when the operating voltage is reversed.  $P_{elec}$  is the electrometer correction factor for the response of the electrometer to the connected ion chamber.

Then the corrected ionization reading, M is converted to dose in the water at the point of reference by using Eq. 14:

$$D_{W} = M.N_{DW}k_{0} \tag{14}$$

Where  $k_Q$  is called the quality conversion factor which depends on the type of ion chamber and the energy of the radiation.

Combining Eqs. 13 and 14;

$$D_{W} = M_{raw} P_{TP} P_{ion} P_{pol} P_{elec} N_{D,W} k_{Q}$$

$$\tag{15}$$

If we assume that temperature and pressure are constant throughout the experiment, then Eq. 15 becomes;

$$D_{W} = M_{raw}.Constant$$
(16)

This is because,  $P_{ior}P_{poP}P_{elec}N_{D,W}$  and  $k_Q$  all depend on the ion chamber, electrometer's operating voltage, and radiation energy, which were constant throughout the experiment. Therefore, during this experiment, we could use the direct ion chamber reading  $(M_{raw})$  as an indicator for radiation dose.

# Experiments

# Effect of contrast material

The dwell times were recorded for different contrast concentrations as described previously. Then the data were normalized to 0% contrast. At the same time, readings from the 0.6 cc Farmer ion chamber were recorded and normalized to 0% contrast concentration. The results are summarized in Figure 10 and Tables 2 and 3.

# Effect of breast air interface

The dwell times were recorded for different over-lying water

thicknesses and the data were normalized to over-lying water thickness d, where d is equal to 7 cm. At the same time, the 0.6 cc Farmer ion chamber was used to measure the dose at 5 cm from the MammoSite<sup>®</sup> catheter axis for different over-lying water thicknesses, and normalized to d equal to 7 cm (full scatter state). The results are summarized in Figure 11 and Tables 4 and 5.

# **Discussion and Conclusions**

#### Effect of contrast material

Both of these experiments were carried out using the same MammoSite<sup>®</sup> balloon with radius of 4 cm. During the first experiment,

Contract	Dwell tir	Normalize to 0%		
concentration (%)	Reading 1 (Sec.)	Reading 2 (Sec.)	Average (Sec.)	concentration
0	266.7	266.7	266.7	1.000
10	268.8	268.8	268.8	1.008
20	268.8	268.8	268.8	1.008
30	268.8	268.8	268.8	1.008









**Figure 10:** Comparison of the normalized dose as a function of contrast concentration for the MammoSite balloon placed in cubical water phantom, at 5 cm from catheter axis. a) Delivered dose measurements from the Nucletron Oncentra treatment planning system. b) Measured dose measurements from the 6 cc Farmer chamber.

the contrast was changed from a concentration of 0% to 30% while keeping the same phantom setup. As shown in Table 3, we observed a prescribed dose reduction from 1.7% to 6.2% at 5 cm away from the balloon axis relative to the dose obtained without contrast material. The largest dose reduction occurred when contrast concentration was at its greatest. According to Table 2 and Figure 10, we observed the predicted dose from the treatment planning system was constant with the different contrast concentrations. Therefore, we can conclude that treatment-planning system does not take into account dose perturbation as a function of different contrast concentrations. Dose perturbations due to contrast concentration have minimal effect on the MammoSite<sup>®</sup> dosimetry as long as the concentration is below 20%.

During the second experiment, we changed the over-lying water thickness from 0 cm to 7 cm while keeping the other parameters the same. We also assumed that 7 cm of over-lying water thickness is more than sufficient for full scatter. As shown in Table 5, we observed a prescribed dose reduction from 0.21% to 7.7% at 5 cm away from the balloon axis relative to the dose obtained with full scatter (when d is equal to 7 cm). As shown in Table 4 and Figure 11, we observed that the predicted dose form the treatment planning system was constant with the different over-lying water thicknesses (d). Therefore, we can reconfirm the conclusion that we made earlier saying that the treatmentplanning system does not take into account dose perturbation due to the effect of the breast air interface. According to Figure 11, when the over-lying water thickness d is less than 1 cm, there is a considerable dose reduction to PTV. Therefore, we can recommend that caution should be taken when planning on using a MammoSite® technique to treat lumpectomy cavities close to the skin surface because of the predicted dose may be clinically different than that of predicted.

According to these results, it is important to develop a treatment



Figure 11: Comparison of the normalized dose as a function of over-lying water thickness for the MammoSite® balloon placed in cubical water phantom, at 5 cm from catheter. a) Delivered dose measurements from the Nucletron Oncentra treatment planning system. b) Measured dose from the 6 cc Farmer chamber.

Contract	lon chan	Normalize to 0%		
concentration (%)	Reading 1 (nC)	Reading 2 (nC)	Average (nC)	concentration
0.0	266.7	266.7	266.7	1.000
0.5	266.7	266.7	266.7	1.000
1.0	266.7	266.7	266.7	1.000
2.0	266.7	266.7	266.7	1.000

 Table 3: Recorded ion chamber readings for different contrast concentrations.

Over-lying water thickness (d) (cm)	Dwell ti	Normalize to		
	Reading 1 (Sec.)	Reading 2 (Sec.)	Average (Sec.)	full scatter (d = 7 cm)
0.0	266.7	266.7	266.7	1.000
0.5	266.7	266.7	266.7	1.000
1.0	266.7	266.7	266.7	1.000
2.0	266.7	266.7	266.7	1.000
5.0	266.7	266.7	266.7	1.000
7.0	266.7	266.7	266.7	1.000

 Table 4: Recorded dwell times from the Nucletron Oncentra treatment planning systems for different over-lying water thicknesses.

Over-lying water	Dwell ti	Normalize to		
thickness (d) (cm)	Reading 1 (nc)	Reading 2 (nC)	Average (nC)	full scatter (d = 7 cm)
0.0	2.629	2.631	2.630	0.923
0.5	2.684	2.683	2.684	0.942
1.0	2.696	2.696	2.696	0.947
2.0	2.773	2.793	2.783	0.977
5.0	2.842	2.841	2.842	0.998
7.0	2.845	2.850	2.848	1

Table 5: Recorded id	on chamber readings	for different over-lying	g water thicknesses.

planning system that performs heterogeneity correction for breast air interface, different contrast materials and different contrast concentrations. During this investigation, we only considered two variables. Therefore, to quantify the uncertainties linked to the Mammo Site<sup>®</sup> brachytherapy technique completely, more research needs to be undertaken regarding the uncertainties due to the balloon deformation and source position [8].

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