Evaluation of the accuracy and efficiency of the in-vivo dosimetry systems for routine cancer patient dose verification

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Key words: In vivo dosimetry, Uncertainty in experimental measurements, Calibration, TLD dosimetry, Semi-conductor (diode) dosimetry.

ISSN: 2070-254X

Abstract

In external beam radiotherapy quality assurance is carried out on the individual components of the treatment chain. The patient simulating device, planning system and linear accelerators are tested regularly according to set protocols developed by national and international organizations. Even though these individual systems are tested errors that can be made in the transfer between systems. The best quality assurance for the system is at the end of the treatment planning chain. In-vivo dosimetry measures the dose to the target volume through indirect measures at the end of the treatment planning chain and is therefore the most likely method for picking up errors which might occur earlier in the chain. In vivo dosimetry, using diodes or thermoluminescent dosimeters (TLDs) is performed in many radiotherapy departments to verify the dose delivered during treatment. The limitation of this technique is that dose can only be in system readout difficulty and type of readout (TLD system and diode) as the patient dose is directly measured. Several authors have investigated the measurements was 1.3%, with a standard deviation of 2.6%. Results were normally distributed around a mean as -0.39 and 0.34 respectively. After the evaluation of in vivo dosimetry brain case as an example, the mean doses for both eyes were 1.8%, with a standard deviation of 2.7%. These results are similar to studies conducted with diodes and TLD's. From these results we can conclude that the diode is superior to TLD, since the diode measurements can be obtained on line and allows an immediate check. Other advantages of diodes include high sensitivity, good spatial resolution, and small size, simplicity of used.

Introduction

Based on the steepness of dose-response relationships, both for local tumor control and for normal tissue complications, an accuracy requirement of 3.5% or one standard deviation (1SD) in dose delivery in radiotherapy daily clinical practice has been formulated.

However, systematic errors in dose delivery for an individual patient can arise, due to: (i) incorrect linac calibration, machine output and field flatness, use of beam modification devices, (ii) incorrect treatment planning system (TPS) calculations, and/or (iii) incorrect patient setup and internal organ motion. Therefore, several international organizations recommend performing in vivo dose measurements.

Currently, the most diffused in vivo dosimetry method is based on the use of two diodes positioned at the beam central axis entrance and exit, respectively, on patient skin surface. Thus the patient midpoint dose, dmax, along the beam axis can be determined by a simple relationship and readings of calibrated diodes. However, this method requires: (i) periodic diode recalibrations; (ii) accurate positioning of the detectors on the patient for every gantry angle; (iii) corrections for photon fluence perturbation; (iv) corrections for temperature, angle of beam incidence and beam energy. Moreover, this method has some limitations when a patient presents asymmetric inhomogeneties along the beam central axis. Experimental and clinical evidence shows that small changes in the dose of 7% to 15% can reduce local tumor control significantly.

So the International Commission on Radiological Units and Measurements (ICRU) recommends that the dose delivered to a tumor be within 5.0% of the prescribed dose [1]. Each of the many steps in the treatment planning and execution will contribute to the overall uncertainty in the dose delivered. Therefore, some organizations (AAPM [3,4], ICRU [1,2]) recommend that in vivo dosimetry (i.e. assess the dose directly in the patient) should be made. In vivo treatment verification includes geometrical and dosimetrical verification. The geometry, i.e. the patient anatomy and tumor location, can be obtained by using a simulator, CT or MRI. Usually the CT and/or MRI data (image fusion) are used to design the 3D treatment plan with a computer treatment planning system. However, due to setup errors and internal organ motion, the planned high dose volume may not agree with the target very well.

Researchers have been working on this, and a new real-time tracking system was introduced. The dosimetric treatment verification is also very important. Each step can contribute to the final dose uncertainty, for example, geometry errors mentioned above, errors introduced by transferring treatment data from the treatment planning system or simulator to the accelerator, errors of beam setting, etc. The final accuracy of the dose delivered can only be checked directly by means of in vivo dosimetry. The most commonly used detector types for in vivo dosimetry are diodes and TLD. In vivo dosimetry is applied to assess the delivered dose to critical organs or in complex geometries where the dose is hard to predict from the treatment. In vivo dosimetry can also be used to monitor the dose delivered in special treatment techniques like Total Skin Irradiation (TSI)

and Total Body Irradiation (TBI).Additionally, in vivo dosimetry is strongly recommended.

The role of in vivo dosimetry in radiation therapy is two-fold:

1. Verify the calculations of the TPS at interfaces, i.e. close to the skin.
2. Evaluate the target dose in order to verify the treatment delivery process.

Materials and Methods

Current in-vivo dosimetry techniques are investigated to find the technique that would best suit Diodes dosimetry incomparsion with TLD system for external beams. These investigated techniques include entrance dose, as well as, entrance and exit dose combined techniques. The diode can be placed at the exit point. Theoretically exit measurements can check all of the parameters mentioned above for entrance measurements, plus changes in patient thickness, contour errors and problems with CT data transfer or CT miscalibration (inhomogeneties in tissue). However, there are some reasons for avoiding the exit position measurements; there are much better more direct methods than in vivo diode measurements to provide quality assurance checks for CT and treatment planning system. These quality assurance methods should be applied long before an in vivo diode measurement is made [5].

Test included Absolute dose:

(1) Beam entrance.

(2) Dose per pulse dependence.

(3) Field size dependence.

(4) SSD dependence.

(5) Energy dependence.

(6) Temperature dependence.

(7) Directional dependence.

(8) Sensitivity as a function of dose per pulse.

A. Diode dosimetry step

(1) Determine the dose at the d_{max} on the central axis using a calibrated ion chamber. For convenience the phantom is usually a plastic phantom. For this work solid water phantom (PTW sided window $30cm^3$) was used. Usually the reference setup is a Gantry of 270 degree, SSD of 100 cm, field size of 10x10 cm^2 (1.00cGy / MU at d_{max}).

(2) With the same setup, tape the diode on the top of the phantom and also on the central axis of the beam. The internal build-up in the diode should be sufficient to absorb electron contamination, and provide electron equilibrium. Diodes should be positioned with the flat surface on the phantom and the build-up side facing the beam. Measure the diode reading for the same irradiation as in step (1).

(3) The calibration factor can be obtained by finding the ratio of the readings from the ion chamber and the diode. This is done automatically by the IVD software. The Diode System used in Current study is SunNuclear Model N-type.

(4) Using this ratio, the diode has been calibrated to read the dose at dmax below the surface, since no inverse square was used to compensate for the slight difference of the diode position.

(5) The calibration factor is verified on a regular basis, because radiation damage affects the diode sensitivity. For p-type diodes, a re-calibration will be necessary after about one cGy. Re-calibration has to be performed much more frequently for n-type diodes due to their faster decrease in sensitivity. Besides a calibration factor, determined under reference conditions, correction factors have to be applied for accurate dosimetry. They originate from the variation in sensitivity of the diode with dose per pulse, the photon energy spectrum, the temperature, and from directional effects.

B. TLD dosimetry step

A total of 40 thermoluminescent dosimeters (TLD) divided into 2 batches (one of 20 and other of 20 TLDs) were used. The thermoluminescent dosimeters are LiF:Mg,Ti (TLD 700) in the form of extruded square ribbons (about 3.1×3.1×0.9 mm^3) manufactured by Harshaw. Thermoluminescent readouts were performed using a Harshaw model 6600 C automatic TLD reader with a linear heating rate of 8 C/s. Nitrogen gas was used. Readouts were taken within 25 s and temperature between 50 C and 250 C. An oven and a furnace were used for annealing procedures of the LiF:Mg,Ti. The annealing procedure used consists of two subsequent annealings: 1 h at 400 C and 2 h at 100°C.

The irradiations were carried out using a Co-60 unit (MSD model Theratronics T780E) with polymethylmethacrylate serving as buildup material (5 mm thick). The reference standard system consists of a cylindrical ionization chamber (Farmer type) model TN30013 (0.6 cm^3) and an electrometer model UNIDOS E T10008 both from PTW-Freiburg. The International Atomic Energy Agency code of practice [6] was followed in the determination of absorbed dose to water.

All TLDs of the 2 batches were annealed and irradiated to same dose. After readout, the procedure was repeated 3 times. A sensitivity factor was determined for each TLD. The intrinsic precision of each batch was evaluated calculating the pooled standard deviations [7]. Supralinearity of response with dose of LiF:Mg,Ti after 1 Gy was investigated by determining the variation of TLD response with doses between 0.25 Gy and 3.5 Gy [7,8].

Results

A. Results and Discussion for Diode

The reproducibility of the sensitivity of the Diodes detector, that is, the change in reading for irradiation to known doses, was measured. The sensitivity was investigated for a range of simulated treatment fraction deliveries from 2-10 Gy/fraction, including simulated breaks for gantry motion between beams. Diodes were irradiated at 1.5cm depth in a 30×30×30cm^3 water slab phantom with a standard Field size 10×10 cm^2 and 6MV photon beam at 100cm SSD. Six field treatment fractions of 2-10 Gy/fraction were simulated by irradiating a diode up to the fraction dose in six equal increments (to simulate irradiation from each beam). The fraction sizes simulated are given as shown in Figure1, showing response of the diode with respect to Monitor unit, and in Figure2 showing response of the diode with respect to field size and is compared to the ionization chamber and Treatment Planning System (TPS) outputs.

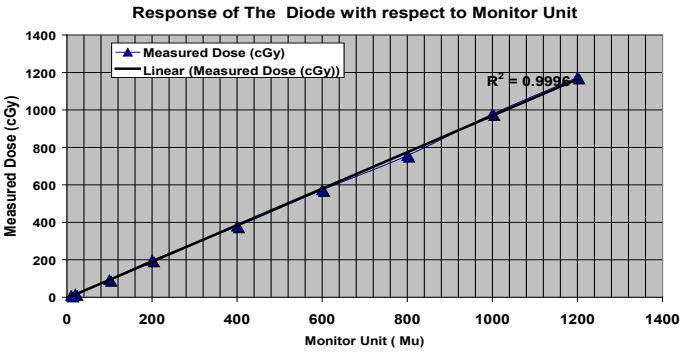

Fig 1: Response of the Diode with respect to Monitor Unit (Mu).

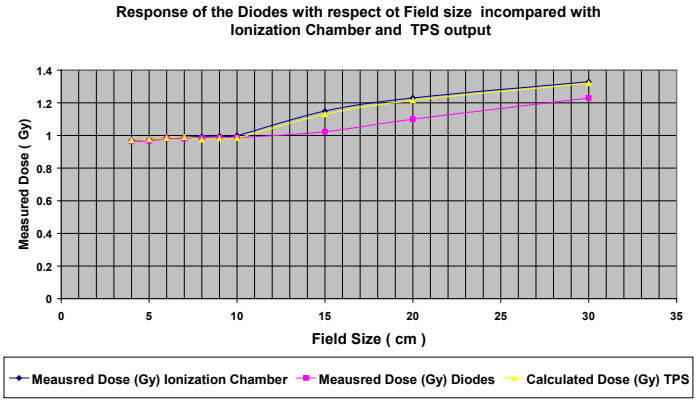

Fig 2: Response of the Diodes with respect to field size in compared with ionization chamber and TPS output.

Diodes correction factor of one type of photo diode (6MV – diode) for 6 MV photon beam as function of Field size for open and different standard wedge angle as shown in Figure (3);

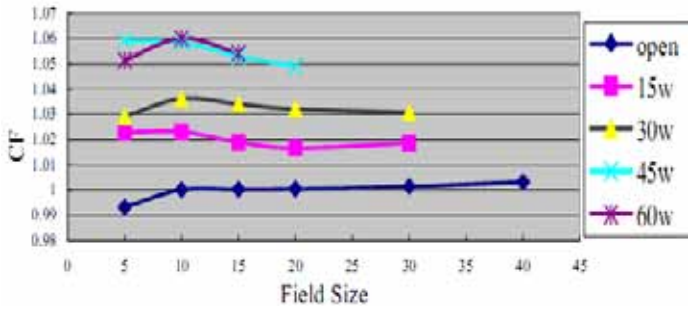

Fig 3: Correction factor of 6MV – diode of as a function of the field size, FS for Open field and different Wedge Angles, for entrance measurements. All data in this figure are for SSD 100 cm and ratio to ionization as standard (for standard Field size 10 X10 cm^2). (Max Sq.field size for W 60 is 15 cm).

As illustrated in Figure (3), the correction factor (CF) of this diode doesn't change much when the field size changes. From this figure, one can see that CF increases with wedge angle. This is because dose per pulse decreases with increase of wedge angle, due to the beam hardening also contributed to this effect. The field size effects are more significant than QED diode. The field size dependences for open, 15 degree and 30 degree wedged fields are almost the same, but those for 45 degree and 60 degree wedged fields are larger, up to 6%. Diodes correction factors of the 6MV IsoRad photo diode as function of SSDs for open and different standard wedged fields is shown in shown in Figure4. Also diodes correction factors dependence on the SSDs of the same type of photo diode with energy of 15MV photon beam is shown in Figure5.

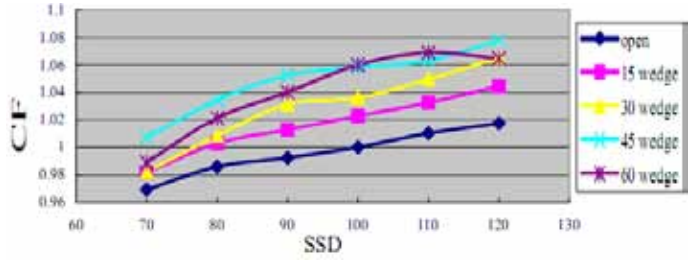

Fig 4: Diode correction factors as a function of the SSD for entrance measurements. All data in this figure are for field size 10x10 cm^2 (6MV photon).

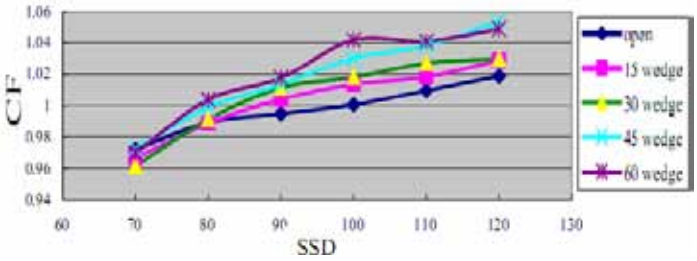

Fig 5: Diode correction factors as a function of the SSD for entrance measurements; (15MV photon). All data in this figure are for field size 10x10cm^2.

The DCF for a wedged field is generally larger than that for corresponding open field, since dose per pulse becomes lower for a wedged field. When the SSD decreases, the number of contamination electrons and scattered low energy photons from head are able to reach the sensitive part of the diode detector so the DCF ratio of ion chamber and diode reading decreases.

B. Results and Discussion for TLD

TLDs are not linear in their glow curve output as the dose increases. The dose is generally Supralinear till a saturation point is reached after which the dose tapers off as all traps are full and are therefore less likely to accept electrons (saturation region).

The useful range of the TLDs is therefore an important factor when measuring dose. TLDs suffer from fade which means that after irradiation they lose electrons in the trapped regions through random processes over time. The fading of TLD is small (5-10% per year for LiF) when used in the in-vivo dosimetry. TLDs have a similar constancy to Diodes on a measurement to measurement basis with typically 1 σ being 2% for radiotherapy applications [9]. TLDs have excellent water equivalence and are less susceptible to low energy radiation than other dosimeters in the 30 to 100 keV range. They also do not require cables for measurement which makes them ideal for mail based studies. TLDs have no dose rate or temperature dependence [8].

Before we use TLD batches measurements for fading has to be done (after eliminating low level signals). 5 sets of Chips (each set contains around 6 chips) wrapped into plastic foil and kept on the bolus material at 1cm depth were used under standard conditions, SSD: 100 cm; FS: 10 x 10 sqcm; Applied dose: 1 Gy and at 1 cm depth. Each set of detectors was read in different time after their exposure. Dosimeter responses versus Time delay (t) are plotted as in Figure7. Responses are in cGy.

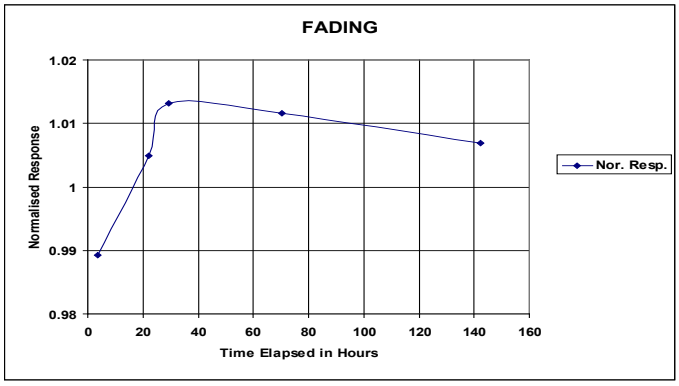

Fig 7: Measurements for Fading (After Eliminating Low Level Signals).

TLD chips were wrapped into plastic foils and kept on the bolus material of size 1.5x30x30 cm^3. SSD: 10x10 cm^2; SSD: 100 cm; Depth: 1cm; 1cm bolus material

was kept on the chips. Sufficient thickness (11 cm) of bolus material and PMMA sheets were kept under the chips for backscatter. Different doses of Co^{60} beam were delivered on the chips at 1cm depth and SSD: 100 cm. The TLD response with dose was plotted versus the dose for each batch. Uncorrected response against the applied dose is plotted as in Figure8. The batch of 20 TLDs was found to have an intrinsic precision of 1.5%. The other batch of 20 TLDs was found to have an intrinsic precision of 1.6%. The thermoluminescent dosimetric system allows individual dose measurements with an expected overall uncertainty lower than 3%. This overall uncertainty is less than 5%, the action level recommended by ICRU [10].

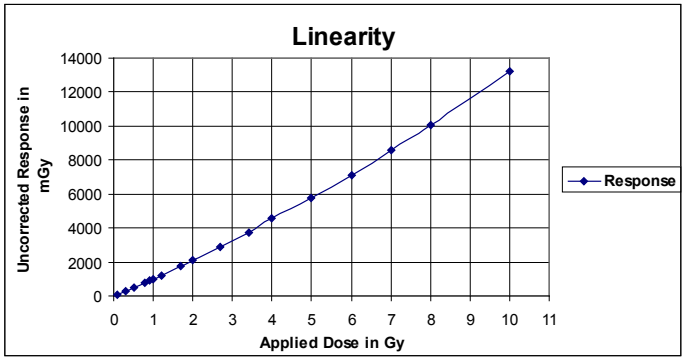

Fig 8: Linearity for TLD patches as function in different doses.

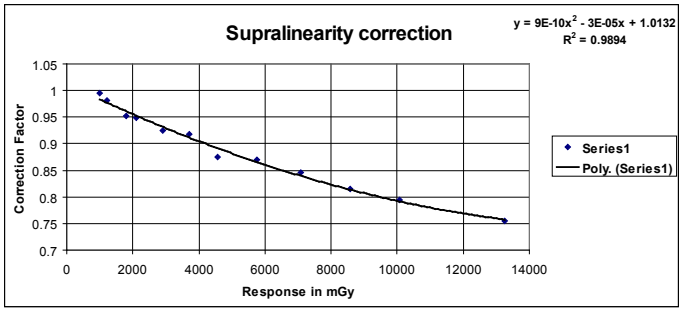

Fig 9: Supralinearity correction for TLD patches as function in different doses in cGy.

A formula proposed by (Mayles et al) [7] was applied to correct for the photon beam of Supralinearity on the TLD response curve. Figure8 show a linear region up to about 1 Gy, from which the TLD response becomes supralinear as in Figure9, consistent with the literature [7,8]. The linear is to the experimental data corrected by the formula proposed by (Mayles et al) [7] showed a correlation equal to 1, showing its applicability in clinical practice.

For TLD dosimetry to be accurate, quality assurance of the system should be performed periodically. TLD accuracy depends on the annealing process. The annealing process involves heating in order to remove all trapped electrons in the crystal, so that it is primed for receiving radiation again. During annealing, the linearity of the heating and cooling, as well as heating and cooling time periods affect the sensitivity of the TLD. The TLD should therefore be annealed in the same way every time. The reader heating linearity and time should be kept constant, as should the gain on the Photomultiplier tube (PMT).

Conclusion

TLDs and diodes are used with entrance dose or exit dose measurements to verify the entire planning process to delivery. There are obvious errors associated with exit dose measurement such as detector placement on the exit surface. Measurement error is also harder to trace back to the source of the problem when making exit dose measurements. Entrance dose is far easier to predict. For these reasons entrance dose has been the favorite choice for institutions measuring IVD system [11].

Although the TLDs response presents a good reproducibility, (1.6 ± 0.7) %, on average, the uncertainty in experimental measurements and in the dose values obtained by the planning system achieves about 10%. The TLDs calibration was based on mathematical formulas developed in the AAPM Report 51.

Table 1: The Comparison of the measurements and percentage variation between measured dose by the Diodes, TLD , ionization chamber and calculated dose for standard field size by TPS (Eclipse Version 10.0).

Mu	Diode	TLD	Ionization Chamber	TPS
100	99.3	99.47	100	100
200	198.6	198.93	200	100
300	**297.0**	**298.00**	**300**	**300**

- Mean variation between measured dose with diode and TPS was found to be 0.7 % (SD 0.36)
- Mean variation between measured dose with TLD and TPS was found to be 0.53 % (SD 0.32)

Table 2: Comparison between vivo dosimetry for head phantom for different detectors like TLD, Ionization chamber (IC) and Diode system all data are compared with data for planning system.

Location	IC	Detector 1	Detector 2	Mean
Lt Lat Field center[a].	45.4	1505.965	1562.368	1534.167
Rt Lat Field center[a].	47.1	1524.651	1563.146	1543.899
Lens of Lt. Eye.	1.3	21.687	22.331	22.009
Lens of Rt. Eye	1.3	23.884	24.409	24.1465

Location	% of Dose	TLD For 50.4Gy	Diode	TPS
Lt Lat Field center[a].	85.23	42.96	45.20	46.0
Rt Lat Field center[a].	85.77	43.23	43.31	47.0
Lens of Lt. Eye.	1.22	0.62	0.634	1.1
Lens of Rt. Eye	1.34	0.68	0.691	1.1

[a] TPS based plan: Planned for 6MV beam: Standard Head phantom, Parallel Opposed Lateral Skull Fields; Open and 15 wedge beams for both Rt and Lt lateral fields, dose of 180 cGy/fraction is delivered to normalization point in 28 fractions. All dose values are in cGy.

The results of in vivo entrance dose measurements are presented in table 2 and showed a mean percentage deviation of measured dose from expected dose of 99% with a standard deviation of 2.6%. The comparison between the standard deviation of the mean percentage deviation of measured dose from expected dose (2.6%) and the estimated overall uncertainty of individual dose measurements (3%) indicates that small discrepancy between the measured and expected mean value (-1%) was due to limitations of the dosimetric system. In this pilot study no discrepancies larger than 5% between the expected dose and measured dose were detected. These data are argument with different published data [7].

The diode is superior to TLD, since the diode measurements can be obtained on line and allows an immediate check. Other advantages of diodes include high sensitivity, good spatial resolution, small size, simple instrumentation, no bias voltage, ruggedness, and independence from changes in air pressure. The sensitivity relative to the ionization volume is high for a semi-conductor, about 15,000 times higher than for an air ionization chamber.

The average energy required to produce an e-hole pair in silicon is only 3.5eV compared with 34eV in air. The sensitive volume can thus be small, and hence the diode detector has high spatial resolution. However, there are many factors that can affect the response of the diode to radiation, and diodes are different from one to another, even from the same batch, same model and same manufacturer. So the commissioning or characterization of every diode individually is necessary for accurate dosimetry [12].

Radiation damage induces recombination centers in the crystal lattice resulting in a greater chance of recombination of charges thereby reducing the resulting current. Sensitivity therefore is reduced with dose. At higher dose rates recombination centre become "occupied" this means that there is less recombination and that more current flow an over-response will therefore become apparent with high instantaneous dose rate. This effect is more pronounced in n-type diodes than in p-type diodes [13].

Diodes have a proven track record for giving in-vivo dose with a low intrinsic error. IVD measurement error for diodes and TLDs is similar to diodes that have an intrinsic error of 2.0% [14] and TLD about 4.9% [15]. Another factor influencing choice of dosimeter is man-hours per readout. TLDs require the greatest amount of time per readout, while Diodes require shorter times, as preparation of these detectors consists of placing detector on the patient and pressing a button once the initial calibration of the relevant factors has been made.

Acknowledgments

All the authors thank the anonymous reviewers for their helpful comments on the original manuscript. This study was supported by grants from Al-Hosain Hospital, Faculty of Medicine, Al Azhar University and Tanta University.

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